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Quality Assurance and Patient Dose Monitoring Methods in Computed Tomography

Master's Thesis
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Tietokonetomografiassa hyödynnetään ionisoivaa säteilyä, joka voi vahingoittaa solujen perimää. Tästä syystä potilaiden saamien säteilyannosten seuraaminen ja ALARA-periaatteen noudattaminen on tärkeää. Tietokonetomografialaitteen toiminta varmistetaan asiaankuuluvalla laadunvarmistuksella, joka pitää sisällään laajan kattauksen erilaisia laadunvarmistustestejä.

Tässä diplomityössä on tutkittu laadunvarmistuksen ja potilasannosten seurannan menetelmiä tietokonetomografiassa. Työ on tehty lähettämällä kysely valikoiduille sairaaloille sekä vierailemalla näistä osassa kattavamman näkökulman saamiseksi.

Kysely sisältää kysymyksiä laadunvarmistuksesta ja potilasannosten seurannasta. Laadunvarmistuksen osalta tavoitteena on kartoittaa, millaisia testejä sairaaloissa tehdään, kuka niitä tekee ja kuinka usein sekä miksi juuri kyseiset testit ylipäätään tehdään. Lisäksi tavoitteena on selvittää, miten ja kenen toimesta potilasannoksia seurataan ja kuinka niihin liittyvät hyväksyttävyysrajat ovat määräytyneet. Kuudesta ennalta valikoidusta sairaalasta neljän vastaukset on mukana tuloksissa. Aikataulusyistä vierailu tehtiin lopulta vain kahteen sairaalaan.

Vastausten perusteella sairaalat seuraavat lainsäädäntöä ja säännnöksiä hyvin. Suoritettavat laadunvarmistustestit ovat pääsääntöisesti määräaikaistestejä, joiden lisäksi laitteen asennuksen yhteydessä tehdään tiettyjä lisätestejä. Sairaalat ovat pääsääntöisesti tyytyväisiä nykyisten laadunvarmistustestien kattavuuteen, vaikka joitakin kuvanlaatuongelmia on ilmennyt. Potilasannosten seurantaan sairaalat joko käyttävät annosseurantaohjelmaa tai ovat aikeissa hankkia sellaisen lähitulevaisuudessa. Potilasannosten raja-arvoissa seurataan pääsääntöisesti STUK:n asettamia rajoja, mutta tietyissä tapauksissa käytetään omia sairaalakohtaisia rajoja. Jatkotutkimuksena kysely tulisi lähettää useammalle sairaalalle, joskin kysymyksiä tulisi muokata jo saatujen vastausten perusteella.

Asiasanat:	CTDI, Laadunvalvonta, Potilasannos, Säteilyannos, TT, Tietokonetomografia
Kieli:	Englanti

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ABSTRACT OF
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CT utilizes ionizing X-ray radiation which, can harm the genotype of living cells. Therefore it is important to monitor the patient radiation doses and follow the ALARA principle. The operations of a CT scanner are ensured by adequate quality assurance program, which includes a variety of quality assurance tests.

In this thesis the quality assurance and patient dose monitoring methods in CT are studied. The study is done by sending a survey to selected hospitals, and additionally, visiting the hospitals for a more comprehensive perspective.

The survey includes questions about quality assurance and patient dose monitoring. The goal with quality assurance is to map out what kind of tests the hospitals perform, who conducts them, and how often and why they are performed. The aim is also to find out how and by whom the patient doses are monitored, and how the acceptable levels have been determined. From six selected hospitals four are finally included in the results. From these four hospitals only two were visited due to scheduling issues.

Based on the answers, the hospitals follow the legislation and regulations well. The performed quality assurance tests are mainly periodic tests even though some additional acceptance tests exist. Despite the fact that some image quality problems have been encountered the hospitals are overall satisfied with the current extent of the quality assurance tests. When it comes to patient dose monitoring hospitals are either already using a dose management software or are acquiring one in the near future. The acceptance levels given by STUK are mainly followed, however, own local levels are used in certain situations. In the future more hospitals should be included in the survey and questions could be modified based on current results.

Keywords:	CT, Computed tomography, CTDI, Patient dose, Quality assurance, Radiation dose
Language:	English

Acknowledgements

Finally my thesis is done and my studies are coming to an end. All in all, it was a bit longer journey that I had anticipated but at this point it really doesn't matter.

First of all I want to thank my supervisor Prof. Ari Koskelainen, who has been a great support not just during this thesis but also otherwise in my studies by helping and guiding me on getting all the necessary pieces for my decree to completion.

This thesis was made for Siemens Healthcare and I want to give my gratitude for the opportunity. I also want to thank all my great colleagues for the years we worked together, especially the great women in CCC team. Finally, special thanks goes to my advisors Mats Lindholm and Antti Laine for their guidance, without you the thesis would not have finished.

My studies and life have been strongly influenced by student voluntary work. Many times it has seemed that volunteering has been the main thing, not the studies itself. Ever since I started my studies as a freshman in The Guild of Physics in 2009 I have experienced unforgettable memories that I will cherish for the rest of my life. I have had the privilege of meeting incredible people and making friendships that will last a lifetime. I want to thank all of you who have been a part of this Teekkari-life of mine, you have taught me so much. Special thanks goes to UltrabRaati12, ITMK12, ITMK13, TJ13, TJ14, Suurhylkeet, Fyysikkospeksi, Lunnit and Suokanat.

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I wrote the thesis mostly at home so I spent most of my days without decent human contact. Luckily we happen to have two adorable furballs that kept me company and stopped me from falling to total insanity. So as

a proper #crazycatlady I want to thank our cats Maxwell and Boltzmann (Figure 1).

Figure 1: Maxwell (right) and Boltzmann doing their part of this thesis.

Last, but definitely not least, I want to thank my beloved Mikko. You have been the greatest support on those moments it felt this thesis would never be finished and have given me the strength to keep on going. I love you.

Tekniikan kehdossa, 4. kesäkuuta 2018

Riina Kärnä

Abbreviations

ALARA	As Low As Reasonably Achievable
ART	Algebraic Reconstruction Technique
BP	Back Projection
CBCT	Cone-beam Spiral Computed Tomography
CT	Computed Tomography
DAS	Data Acquisition System
DLP	Dose Length Product
EBCT	Electron Beam Computed Tomography
ED	Effective Dose
FBP	Filtered Back Projection
FOV	Field Of View
FWHM	Full Width Half Maximum
FWTM	Full Width Total Maximum
HUS	Helsingin ja Uudenmaan Sairaanhoitopiiri (The Hospital District of Helsinki and Uusimaa)
HVL	Half Value Length
IRS	Image Reconstruction System
KYS	Kuopion yliopistollinen sairaala (Kuopio University Hospital)
MSCT	Multi-slice Spiral Computed Tomography
OYS	Oulun Yliopistollinen Sairaala (Oulu University Hospital)
PHKS	Päijät-Hämeen Keskussairaala (Päijät-Häme Central Hospital)
QA	Quality Assurance
RF	Radio Frequency
SJKS	Seinäjoen Keskissairaala (Seinäjoki Central Hospital)
SSDE	Size-specific Dose Estimate
SSCT	Single-slice Spiral Computed Tomography
STUK	Säteilyturvakeskus (Radiation and Nuclear Safety Authority)
TYKS	Turun Yliopistollinen Sairaala (Turku University Hospital)

Acronyms

A	Anterior-posterior dimension of a scanned body part [cm]
c	Speed of light, $c = 3 \times 10^8\ m/s$
$CTDI$	Computed Tomography Dose Index [mGy]
$CTDI_{100}$	Computed Tomography Dose index for 100mm long ionization chamber
$CTDI_c$	Computed Tomography Dose Index at the center of a phantom
$CTDI_p$	Computed Tomography Dose Index at the periphery of the phantom
$CTDI_{vol}$	Volume Computed Tomography Dose Index
${CTDI_{vol}}^{16}$	Volume Computed Tomography Dose Index for $16cm$ diameter reference phantom
${CTDI_{vol}}^{32}$	Volume Computed Tomography Dose Index for $32cm$ diameter reference phantom
$CTDI_w$	Weighted Computed Tomography Dose Index
d	Thickness of material [m]
D	Effective diameter
$D(z)$	Radiation dose profile along the z-axis
E	Energy of an X-ray photon $[J]$
${f_{size}}^{16X}$	Correction factor for a $16cm$ diameter reference phantom
${f_{size}}^{32X}$	Correction factor for a $32cm$ diameter reference phantom
h	Planck's constant $[Js]$, $h = 6.63 \times 10^{-34}\ Js$
I	Outgoing x-ray beam intensity
I_0	Incoming x-ray beam intensity
$I(x)$	Intensity profile
K_1	Radiation dose
K_2	Radiation dose
l	Scan length in z-direction [cm]
L	Lateral dimension of a scanned body part [cm]
n	Number of detector rows

Q_1	Electricity
Q_2	Electricity
S	Sum of anterior-posterior and lateral dimension
v	Frequency $[Hz]$
κ	Organ specific constant
λ	Wavelength [m]
μ	Linear attenuation coefficient [m^{-1}]
μ_{air}	Linear attenuation coefficient of air, $\mu_{air} = 0$
μ_m	Mass attenuation coefficient
μ_{water}	Linear attenuation coefficient of water
ρ	Mass density
δz	Detector width in z-direction

Contents

Chapter 1

Introduction

1.1 Background

Today X-ray computed tomography (CT) is an essential part of radiological diagnostics [1]. CT is a non-invasive imaging method which allows us to obtain a clear anatomical image without violating the outer surface of a person's body [2]. CT is based on X-rays discovered by Wilhelm Conrad Röntgen over 100 years ago at the end of the 19th century [3]. Generally, in X-ray imaging a beam of X-rays passes through a tissue, or other media, while experiencing insignificant scattering. The analysis of the transmission data of the X-ray beam are the basis of X-ray imaging.

With conventional X-ray imaging the information from a three-dimensional body is projected and flattened onto a two-dimensional plane [4]. This process may lead to the loss of subtle irregularities by overlapping tissue structures. For example, within the lung the convoluted shapes of the the larger air paces and the ribs can can cause strong variations in the X-ray beam attenuation leading to obscured soft tissue lesions.

CT tackles this problem by using the same X-rays in order to present cross-sectional images, slices, of the body [4]. For one slice, CT gathers and digitizes two-dimensional X-ray shadows from several perspectives around the patient. By starting from the obtained data set and working backwards, a mathematical reconstruction of the spatial distribution of the X-ray attenuation properties of the materials responsible for the set of images is acquired. The final result obtained from the image sets and several slices is a three-dimensional image of the patient. An example of the difference between conventional X-ray and CT is presented in Figure 2. Detecting cancer is difficult from the plane image but is is clearly visible in the CT slice.

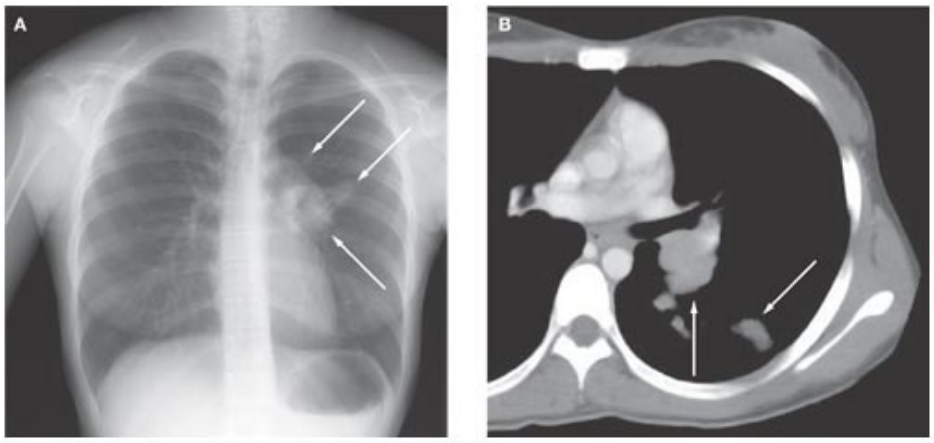


Figure 2: Lung cancer imaged with conventional X-ray (A) and with CT (B). The cancer is more visible in the CT image. [5]

1.1.1 Origins of Computed Tomography

The ideas behind CT can be traced back to the beginning of 20th century. In 1917 Johann Radon wrote a paper about determination of functions from their integral values along certain manifolds (translated by P. C. Parks in 1986) [6, 7]. In his paper Radon proved that the distribution of a material or material property in an object layer can be calculated if the integral values along any number of lines passing through the same layer are known. This has later been referred to the Radon transformation.

CT only became feasible in the sixties with the modern computer technology development. The two generally credited inventors of computed tomography are Allon MacLeod Cormack and Godfrey Newbold Hounsfield [8]. They won the Nobel Prize in Physiology or Medicine in 1979 for their outstanding achievements.

A. M. Cormack was the first one to carry out experiments on medical applications associated to tomography. Between 1957 and 1963 he developed a method for calculating radiation absorption distributions in the human body based on transmission measurements [9, 10]. In 1963 he reported his findings from the investigations of perhaps the first CT scanner actually built. Due to the time and the difficulties in performing the necessary calculations unfortunately little interest was paid to his work at the time [11]. During his work Cormack was unfamiliar of Radon's work and discovered it only later.

The successful practical implementation of the theory was finally achieved by G. N. Hounsfield. He started his work towards the first clinical CT scanner in 1967, and like Cormack, worked without any knowledge of earlier findings

[11]. Hounsfield used the power of computers for complicated calculations to tackle the image reconstruction problem, giving CT its practical expression [2]. In the first scanner, linear scans were performed on a rotating specimen in 1-degree steps [12]. The data acquisition and the image production took nine days to complete due to the use of a low-intensity americium gamma source.

Figure 3: Illustration of the patient position in the first clinically available CT scanner which was installed in Atkinson Morley's Hospital in London in September 1971. The first patient was a woman with a suspected brain tumor. From the obtained images, it was possible to clearly differentiate the physiological areas of the brain and the darker pathological area of the developing tumor. [12]

1.1.2 Ionizing radiation and Radiation Dose

CT utilizes X-ray radiation which is ionizing. Ionizing radiation has high enough energy to release electrons from the atoms of the target material or to break the molecules of the material [13]. Therefore, ionizing radiation can

harm the genotype of living cells [14]. The most harmful health effects can be categorized to deterministic and stochastic [15]. Deterministic health effects are biological, such as harmful tissue reactions, and they produce immediate change. Deterministic effects appear above a high threshold dose, generally after radiation has killed a large proportion of cells [16]. Increasing radiation dose usually increases the severity of the effect. Stochastic health effects are genetic or carcinogenic, as cancer and hereditary effects in offspring, and they develop during a longer period of time. Stochastic effects do not appear on a specific dose threshold, but rather with a probability. The probability is thought to increase along the dose.

Compared to planar radiography, CT scanners create superior high contrast images which is important especially for diagnosis involving soft tissue (organs) [17]. This makes it very useful and is increasingly the technique of choice. However, the patient radiation dose may be significantly higher compared to other imaging modalities. Due to effects of ionizing radiation, this comes to an especially important role in the case of a pregnant patient or a child.

To minimize the radiation dose, it is important to follow the guiding principles for radiation protection [18]. For a right dose, the specific diagnostic task and the specific patient attenuation (patient size) has to be taken into account in CT examination planning. This follows the ALARA (as low as reasonably achievable) principle. It also has to be considered, whether a CT scan is appropriate for the specific individual patient. Therefore, a CT exam should be performed only when the radiation dose is justified by the potential clinical benefit.

A wide terminology for the patient radiation dose evaluation in CT exists. In the past years one of the most commonly used type of dose in reports is the effective dose [16]. For example, the effective doses for the some common CT scans are: $2\,mSv$ for head, $9\,mSv$ for lungs, and $12\,mSv$ for abdomen [19]. For reference, the mean effective dose for a Finn was in 2012 $3.2\,mSv$, from which half, $1.6\,mSv$, was attributed to the indoor radon exposure [20]. Also annually roughly one third of the effective dose, $1.1\,mSv$, is caused by natural background radiation. In 2011, the annual the effective dose caused by medical x-rays was determined to be $0.45\,mSv$. If the effective dose received within under 24 hours reaches $6000\,mSv$ the causes are radiation sickness and they can also be fatal [21].

1.1.3 Quality Assurance

CT scanners are under constant technical development which results in increasing clinical applications [17]. As the scanners get more complex on

operations and applications, a careful monitoring is required. This ensures appropriate examining conditions exist and procedures are optimized for diagnostic quality and patient dose. To obtain adequate monitoring, it is essential to promote and facilitate the implementation of a quality assurance (QA) programme. This programme includes e.g. appropriate training for the personnel, use of well-designed and properly operating equipment, and suitable examination protocols.

In Finland the QA of CT equipments is regulated based on radiation act (592/1991, change 1142/1998, § 40) [22]. The act states that the responsible party shall implement planned and systematic measures to ensure the radiation sources and accessories, and associated equipment are in good condition. The responsible party is also obligated to ensure the instructions and procedures concerning these are appropriate.

The medical use of radiation is regulated by Decree of the Finnish Ministry of Social Affairs and Health [23]. According to the decree (423/2000, § 32), the QA programme must present the main tasks involved in supervising the operating condition and performance characteristics of radiological equipments [24]. The responsibilities and instructions concerning the measures of each individual item of the equipment are specified separately.

The legislation, safety regulations, and guidelines related to radiation and nuclear safety are supervised by the Radiation and Nuclear Safety Authority (Säteilyturvakeskus, STUK) [25]. STUKs mission is to protect people, the society, the environment, and future generations from the harmful effects of radiation. STUK operates on a wide scale on radiation safety and they, for example, regulate the use of radiation in healthcare, industry, research, and training.

1.2 Objectives and Scope

The objectives of this thesis are to map out the quality assurance (QA) and patient dose monitoring methods in Finnish hospitals. STUK gives instructions and guidelines regarding both of these. These guidelines are the minimum demands, but hospitals can have some tests and methods of their own on top of them. The goal is to specifically find out the extent and variety of these own QA tests, and why they are performed. For possible future improvements, it is also of interest to see if some problems regarding CT image quality occurs despite the current QA. With patient dose monitoring the goal is to figure out the extent of dose monitoring and how the acceptable levels have been determined.

1.3 Structure

This thesis has the following structure. Chapter 2 introduces the physical background and principles of CT imaging. First the basics of X-rays and their utilization on CT is presented. The basic principles and the main generations of the CT scanner are then described. Lastly the CT scanner configuration is introduced and followed by a description of the clinical use of the scanner. In Chapter 3 the radiation dose in CT is explained. The main topics are the radiation dose terminology, which is used to follow up the radiation dose in patient, and the factors affecting the dose. The image quality is also presented here since it is closely related to the radiation dose. Chapter 4 introduces the principles and phases of QA in CT. The operation boundaries and the measurement uncertainties give frames for QA and are therefore presented. Lastly the QA instructions and guidelines specially for CT are given and the available tools for radiation dose monitoring are displayed. In Chapter 5 the used methods and in Chapter 6 the results are presented. Finally Chapter 7 includes summary and conclusions for the thesis.

Chapter 2

Computed Tomography

2.1 X-ray Radiation in CT

An X-ray is an electromagnetic waveform with wavelength ranging from a few picometers to a few nanometers. The energy of an X-ray photon E is proportional to its frequency ν and is described as [11]

$$E = h\nu = \frac{hc}{\lambda}, \tag{2.1}$$

where $h = 6.63 \times 10^{-34}\, Js$ is the Planck's constant, $c = 3 \times 10^{8}\, m/s$ is the speed of light, and λ is the wavelength of the X-ray. X-ray photons with longer wavelengths have smaller energies than photons with shorter wavelengths. For convenience, the energy of X-rays is generally expressed with the unit of eV ($1eV = 1.602 \times 10^{-19}\, J$). It is defined as the amount of kinetic energy acquired by a single electron moving across an electric potential difference of one volt.

2.1.1 Generation

X-ray production takes place within matter at the atomic level when a substance is bombarded generally with high speed electrons [11]. The interaction of these electrons and target material leads to different types of collisions. Majority of these involve small energy transfers from a high speed electron to electrons that are knocked out of the atoms resulting to ionization of the target atoms. This leads to delta rays and finally to heat. However, bombarding the substance can generate two types of X-rays: characteristic X-rays and bremsstrahlung (German for braking or deceleration radiation) [2].

Characteristic X-rays appear in transition of electrons between the inner shells of an atom [2]. Each transition of an electron from a higher energy

level shell (L or higher) to a lower energy level shell (K) is accompanied by the emission of a radiation quantum with energy equaling the energy difference between the shells. The quantum has typical energies in the range of $0.052 - 129.544\,keV$ and can only have discrete values since only shells with specific energy levels are allowed. Every time an electron from a higher energy level fills a vacancy on a lower level a characteristic X-ray line is produced. The term "characteristic line" is based on the fact that atoms of different elements are characterized by different sets of lines. A collection of these characteristic X-ray lines are referred as "characteristic series". The generation of characteristic X-rays is presented in Figure 4.

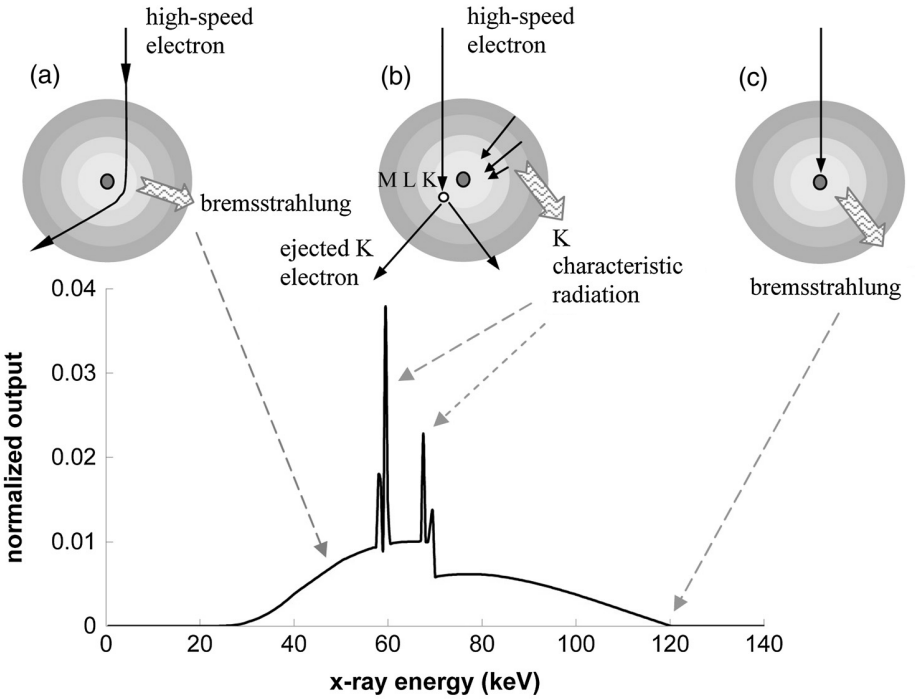


Figure 4: Illustration of X-ray generation types and their relationship to the X-ray energy spectrum. (a) Bremsstrahlung radiation is generated when high-speed electrons are decelerated by the electric field of the target nuclei. (b) Characteristic X-rays are generated when a high-speed electron interacts with a target electron and ejects it. Characteristic X-rays are emitted when a electron from higher electron shell fills the vacancy. (c) The kinetic energy is transformed to X-ray energy when a high-speed electron hits the target nuclei directly. Here tungsten is used as a target material and low-energy X-rays have been filtered. [11]

Bremsstrahlung is deceleration of charged particles caused by electromag-

netic fields within matter [2]. The energy lost in the particle deceleration is emitted as X-ray quanta and can reach any value in a range reaching up to $20\,MeV$. The decelerated charged particle can be an electron, a proton, a particle, or a heavy ion. The generation of bremsstrahlung, also known as continuous X-rays, is presented in Figure 4. Bremsstrahlung is the primary mechanism of X-ray production in diagnostic radiology [26].

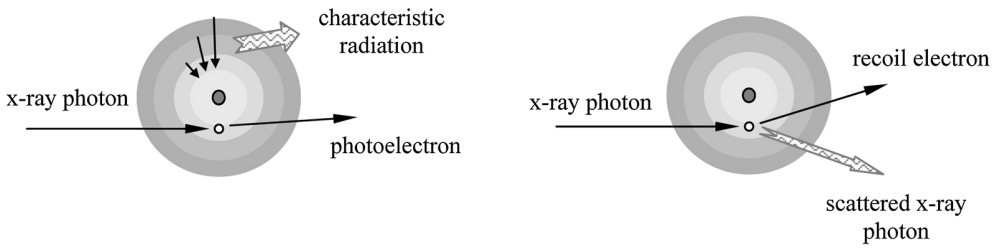
The high-speed electrons bombarding the target anode produce an X-ray spectrum which is the sum of the energies of both the characteristic X-ray radiation and bremsstrahlung [2]. An example of the X-ray spectrum obtained with the X-ray tube is presented in Figure 4. The intensity of the X-rays produced is proportional to the atomic number of the target material and the amount of electrons bombarding it.

2.1.2 Interaction with Matter

The tissue, or other material, has the tendency to remove X-ray energy from the X-ray beam [4]. When a thin beam of X-ray photons passes through matter it becomes weaker and is said to be attenuated as photons are removed from the forward direction of propagation [26]. Attenuation can take place through two processes: scattering and absorption. In scattering the energy is lost by redirecting the photon from the direction of the primary X-ray beam. In absorption the energy is lost by transferring it from the primary beam to the local medium i.e. as heat.

In the photoelectric effect, the X-ray photon annihilates on collision with a bound electron [26]. The photon gives its entire energy to liberate an electron, a photoelectron, from a deep shell of an atom [11]. For detaching the electron the energy of the X-ray photon must exceed the binding energy of the electron. The liberated electron leaves a hole which is filled by an outer shell electron. Since the electron filling the hole is moving from a higher-energy state to a lower-energy state the excess energy generates characteristic radiation. The photoelectric effects results to a photon of characteristic radiation, a photoelectron, and a positive ion (Figure 5a).

In Compton scattering, the X-ray photon also strikes an electron from its shell [11]. However, now the energy of the incident photon exceeds the binding energy of the recoilled electron. As a result the X-ray photon loses part of its initial energy and is scattered or deflected as seen in Figure 5b. The photon deflection angle is from 0° to 180°. Low-energy photons are preferentially backscattered with a deflection angle larges than 90°. High-energy X-ray photons, however, have a higher probability of forward scattering with the deflection angle being less than 90°. Since the deflections angle is wide the scattered X-ray photon does not provide much information of the path and



(a) (b)

Figure 5: X-ray rays are attenuated in tissue by scattering or absorbtion. Scattering can occur by photoelectric effect (5a) or by Compton scattering (5b). Modified from [11]

the individual interactions of the photon with matter. Compton scattering results in a scattered photon, a recoil electron, and a positive ion.

Rayleigh scattering, also referred as coherent scattering, differs from the previous two interactions: no energy is converted into kinetic energy and no ionization of an atom occurs [11, 27]. In this elastic scattering, an X-ray photon is absorbed by an atom and immediately released in the form of a new photon. The new photon has the same energy but is travelling in different direction. Rayleigh scattering occurs mainly at lower energies ($<30\,keV$). X-rays generated for medical CT have a typical energy range of $20 - 140\,keV$, and therefore the Rayleigh scattering does not play a major role. Rayleigh scattering is also referred as coherent scattering.

Pair production occurs only at higher energies when the energy of the photon exceeds $1.02\,MeV$. In pair production the photon is transformed into an electron-positron pair. However, the positron soon annihilates with another electron while creating two photons with the energy of $511\,keV$. [27]

From these X-ray interactions the most significant are the photoelectric effect and Compton scattering [26]. Rayleigh scattering and pair production occur only at extreme energies or outside the radiological field having have only peripheral significance.

2.1.3 Attenuation in Matter

The intensity $I(x)$ of the X-ray beam falls (ideally) exponentially with the thickness of the material it travels [28]. The attenuation rate can be parametrized by the linear attenuation coefficient μ. Alternatively we can determine the exponential attenuation and μ considering the nature of the collisions of individual photons with the atomic electrons of the medium.

Based on the linear attenuation the intensity of the outgoing beam I is

related to the intensity of the incoming beam I_0 by [1]

$$I = I_0 e^{-\mu d}, \tag{2.2}$$

where the linear attenuation coefficient μ is typically expressed in cm^{-1} and the d is the thickness of the material. Now for the linear attenuation coefficient applies

$$\mu = \frac{1}{d} \cdot ln \frac{I_0}{I}. \tag{2.3}$$

The μ is, however, actually a function of both the photon energy and the material [27].

If a single-energy photon travels through a nonhomogeneous medium, (2.2) can be generalized as [1]

$$I = I_0 e^{-\mu_1 d_1 - \mu_2 d_2 - \mu_3 d_3 - \cdots} = I_0 e^{-[\sum_{i=1}^{n} \mu_i d_i]} = I_0 e^{-\int_0^d \mu(s) ds}. \tag{2.4}$$

However, a real X-ray beam contains a whole spectrum of energies instead of single-energy photons and is thus polychromatic. Thus [1]

$$I = \int_0^{E_{max}} I_0(E) e^{-\int_0^d \mu(E,s) ds} dE. \tag{2.5}$$

The greater μ, the more rapidly the beam is absorbed and scattered in the tissue [4]. Respectively, the value of μ depends on the density and effective atomic number of the tissue.

The linear attenuation coefficient can be replaced with the mass attenuation coefficient μ_m [27]:

$$\mu_m = \mu / \rho, \tag{2.6}$$

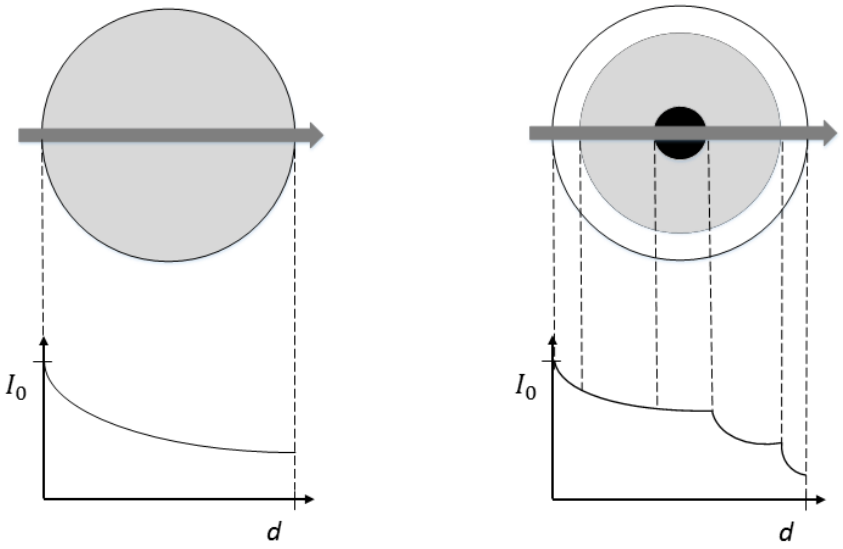
where ρ is the mass density of the attenuating medium.

2.2 Basic Principles of CT

In general, the basic principle of computed tomography is to measure from different directions the spatial distribution of the physical quantity to be examined, and then, based on these measurements, to compute superposition-free images.

2.2.1 Data Acquisition

In CT, the intensity distribution of X-rays is recorded behind the object. To calculate the attenuation value along each ray from the source to the



(a) (b)

Figure 6: Two examples of X-ray intensity attenuation profiles: (6a) homogeneous object and monochromatic radiation, and (6b) inhomogeneous object and monochromatic radiation.

detector, the incoming beam intensity I_0 and the intensity I attenuated by the object need to be measured [1]. Some simple cases of this are illustrated in Figure 6.

In the most simple case (Figure 6a) the object is homogeneous and radiation is monochromatic. Now the intensity falls off exponentially with absorber thickness according to (2.2). If the the thickness d of the object is known, the linear attenuation coefficient μ can be determined directly [1]. However, the distribution of μ remains unknown.

A more interesting case is presented in Figure 6b. Now the to total attenuation, which is resulting from the contribution of each ray path, depends on the local value of the attenuation coefficient μ_i. This can be expressed as the integral over μ along the ray path according to (2.5) [1].

Linear attenuation coefficient is dependent on the energy of the X-ray beam [1]. Therefore, when measuring intensities, the integration is done over all energy intervals as stated in (2.5). In today's CT system this is done automatically. It is sufficient to compute an image to a good approximation with a finite number of measurements of the distribution of the attenuation coefficient $\mu_{x,y}$.

To compute an image with acceptable quality, a sufficient number of attenuation integrals or projection values must be recorded [1] . Therefor,

measurements have to be carried out in all directions, at least over 180°, and several closely spaced data points for each projections must be determined. Today modern CT scanners typically measure $800 - 4000$ projections with $600 - 1800$ data points per projection.

2.2.2 Image Reconstruction

Information on the distribution of attenuation coefficients $\mu_{x,y}$ is given in the form of a set of projection values known as the "Radon transform" of the image [1]. To determine $\mu_{x,y}$ an inverse transformation has to be calculated. Several mathematical reconstruction methods for this are available, for example, algebraic, iterative, two-dimensional Fourier, and filtered back-projection (FBP) [4].

The easiest one to demonstrate is the algebraic reconstruction technique (ART). In ART simple algebra is used to solve a set of linear equations [4]. The simplest case of this is, 2×2 image matrix (Figure 7). With only four pixels, two measurements for two projections yields to four equations and four unknowns which is can be solved easily. When the number of voxels grow ART becomes extremely cumbersome and time consuming. ART was used in the early commercial CTs but is no longer utilized in modern CTs.

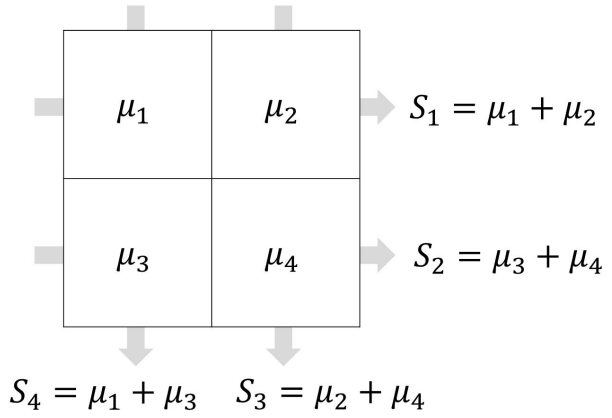


Figure 7: Four-voxel example of ART. Algebraic procedures are the easiest image reconstruction approach to comprehend.

Iterative applications are finding increasingly applications in CT [4]. They can accommodate statistical methods that reduce internal inconsistencies in a dataset and the impact of random noise. For the slice being reconstructed, iteration starts with an educated guess and produces a sequence of improving images. Each of these images is a refinement over the previous one and closer

to the real thing. Iterative reconstruction can also be used to a good image which is already generated with FBP.

For CT, analytical algorithms are faster than iterative ones [4]. The two most common analytical methods are Fourier reconstruction and filtered back-projection. With Fourier, reconstruction calculations can begin only after the completion of all profile measurements, whereas with FBP, the calculations can begin straight after obtaining a single profile. This is a clear advantage for FBP, and therefore, it is the primary reconstruction technique currently used in CT.

FBP is presented with a simple object in Figure 8. FBP starts with an empty image matrix [1]. For simple back-projection, each projection value is added to all picture elements along the direction it has been measured. In addition to the pixel value at a specific image point, each detail in the measured object and presented in the attenuation profile contributes to the whole image. However, the far-reaching signal contributions from the back-projection lead to an unsharp image as seen from the lower left of Figure 8. This is insufficient for complex structure diagnosis.

Unsharpening is avoided by using the reconstruction kernel which convolves each projection before back-projection with a mathematical function [1]. This constitutes a pointwise multiplication of the the convolution kernel and attenuation profile and addition of the resulting values. This presents a high pass filtering procedure which at object boundaries generates over- and undershoots. In case of a positive signal, negative undershoots are generated. This will even out the positive signal contributors outside each object detail. The result for the presented simple object is seen in lower right of Figure 8. The choice and design of the reconstruction kernel affect the image characteristics.

Fourier reconstruction offers a mathematically equivalent procedure for image reconstruction as the FBP [1]. However, FBP is the primary reconstruction technique currently used in CT scanners. [4]

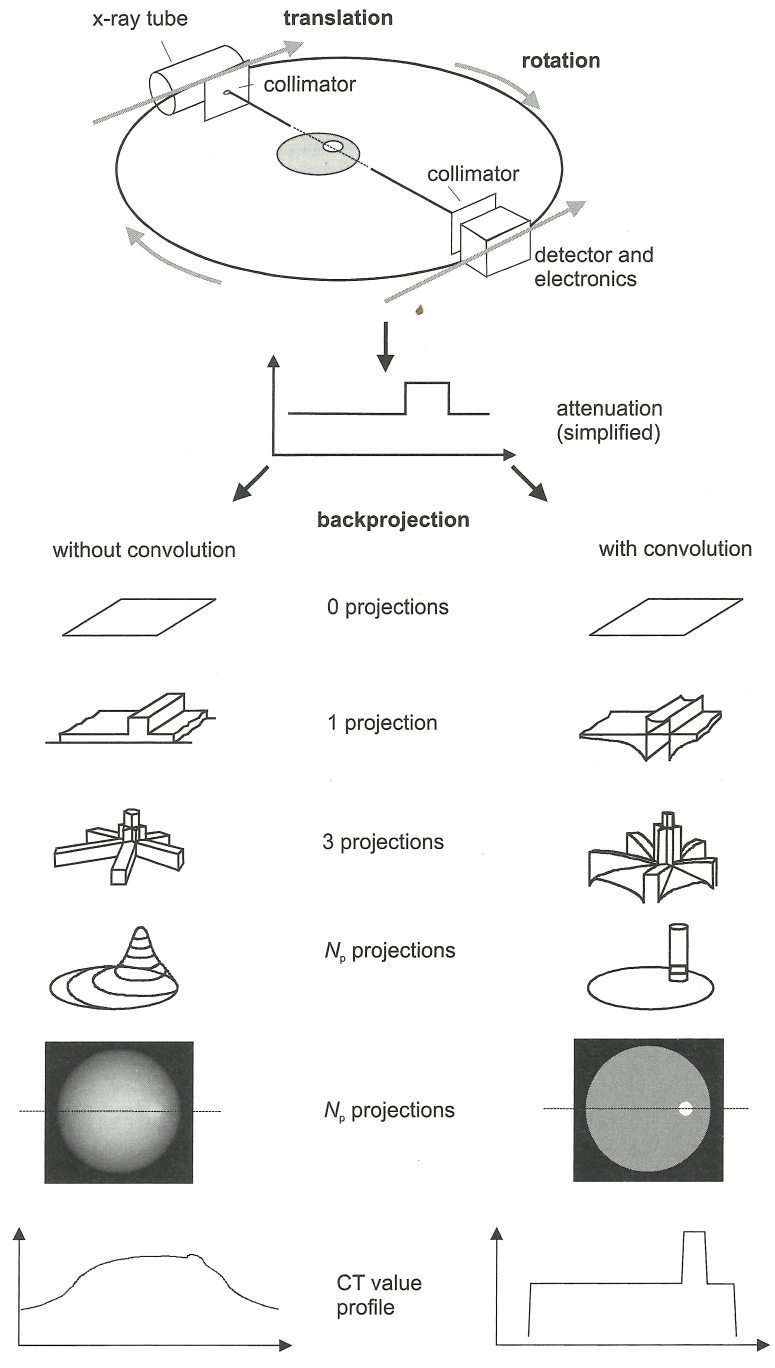


Figure 8: Image reconstructions in CT without convolution (BP) on the left and with convolution (FBP) on the right. Direct back-projection leads to unsharp image. Convolution before back-projection removes the unsharpness. [1]

2.2.3 CT Numbers

The attenuation of any material depends on the energy of the X-ray beam but is not itself generally the direct clinical interest [4]. For simplicity, the attenuation coefficient is removed by normalizing the tissue CT number relative to that of water. The CT number, expressed in Hounsfield units (in HU), of a tissue at a certain point is defined in terms of linear attenuation coefficients as [4]

$$CT\ number = \frac{\mu - \mu_{water}}{\mu_{water}} \cdot 1000, \tag{2.7}$$

where μ_{water} is the linear attenuation coefficient of water at the effective energy of the beam. By definition, the CT number of water is zero. Since linear coefficient of air is zero ($\mu_{air} = 0$), the CT number for it is $-1000\,HU$. CT numbers can range up to a maximum of $3000\,HU$ for bone [2]. Typical approximated CT numbers are presented in Table 1.

Table 1: Approximated CT number values for various materials for an X-ray beam in the range of $80 - 140\,kVp$. [1, 4]

Tissue	**CT number (HU)**
Air	-1000
Lung	-(550 - 950)
Fat	-(80 - 100)
Water	0
Muscle	10 - 40
Kidney	20 - 40
Blood	40 - 60
White matter	46
Gray matter	43
Liver	50 - 70
Spongious bone	50+
Dense bone	1000+

In a single view it is not possible to evaluate or differentiate the whole CT number range (12 bits/pixel corresponds $2^{12} = 4096$ shades of gray [29]) [1]. CT is also widely used to examine soft tissues that have close CT numbers. Therefore, it is desirable that most of the gray scale variation of the display corresponds to soft tissues. For this a procedure called windowing is used: values above the chosen window will be displayed white and as black below it. Windowing is carried out on the CT console by only adjusting the center and width of the window.

2.3 Seven Generations of CT

The CT scanners have continuously been evolving since the early 1970s when the first scanner was introduced [4]. The evolution of CT scanners has had a few radical distinguishable changes that separate scanners to seven generations. At least two of these generations incorporate helical motion and multi-slice capacity. Today nearly all new scanners are helical and multi-slice.

2.3.1 First Generation Scanners

The first CT scanner was built for head imaging by Hounsfield. The first generation scanners use a parallel-beam projection system and can be referred as pencil beam or translation/rotation single detector scanners [1].

The principle for obtaining images with the first generation scanner is presented in Figure 9a. The X-ray tube and the detectors are placed rigidly on opposite ends of the gantry and the beam points always directly to the detectors [4]. When rotating the gantry about the isocenter line the positions of the tube and the detectors are fixed relative to one another. Together they can do both linear and rotational motion relative to the patient. The lateral movement makes a single projection and a rotational movement gathers all the projections necessary to construct an image [1].

The acquisition of individual projections can be either continuous or discrete [11]. However, each projection is obtained only at discrete angle rotation of the projection system. This method is not fast since the X-ray tube and the detector have to travel twice the distance equal to the diameter of the gantry opening during each projection. The scan and image reconstruction are done simultaneously and they both take five minutes [1]. The total examination takes 35 minutes.

2.3.2 Second Generation Scanners

The second generation scanner, presented in Figure 9b, was introduced around 1972 [4]. The principle is quite the same as with the first generation scanner. The only difference is that it uses a fan-shaped beam which enables the projections to cover a larger area of patients body at any time. This reduces the amount of required scan angles thus reducing the scan time considerably to under $20\,s$ [11]. The second generation scanner can be considered as a transitional stage between the parallel-beam and fan-beam systems.

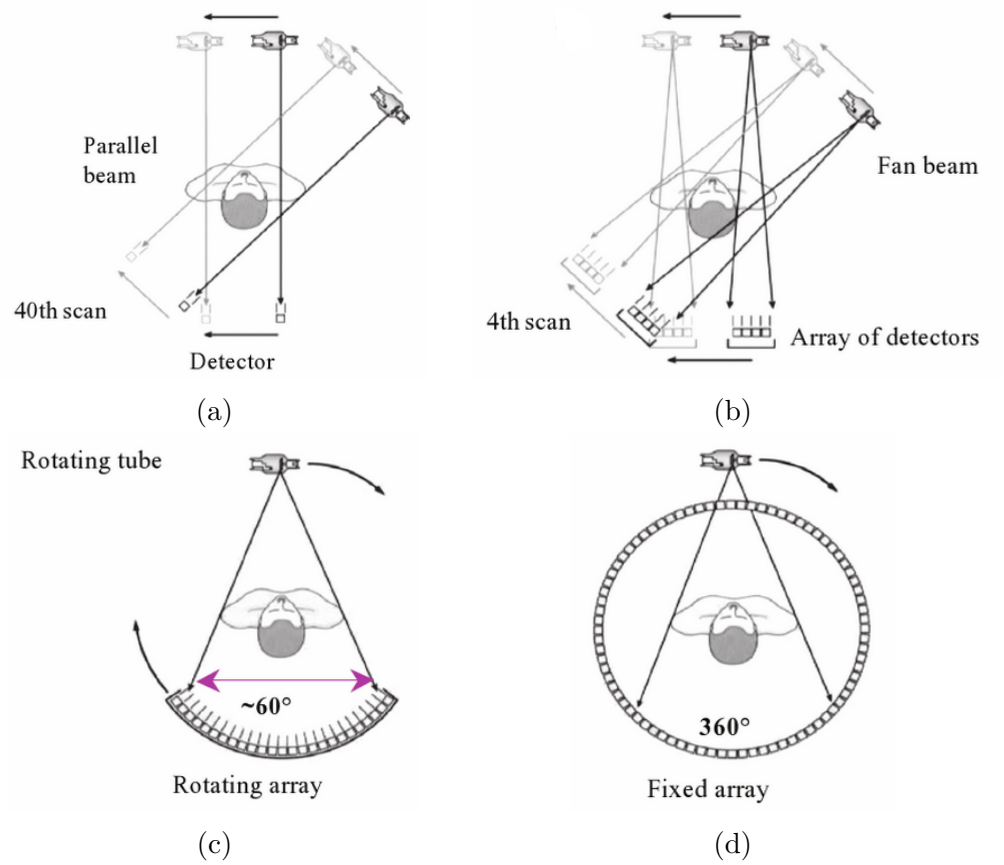


Figure 9: First four CT scanner generations: (9a) first generation scanner (translate-rotate), (9b) second generation scanner (translate-rotate with a fan-beam), (9c) third generations scanner (rotate-rotate), and (9d) fourth generation scanner (rotate-fixed). [4]

2.3.3 Third Generation Scanners

The third generation scanner, introduced in 1976, is presented in Figure 9c [4]. It eliminates the need for linear motions of the X-ray tube and the detectors across the patient. This generation scanner has a fan-beam wide enough to completely cover the patient. To intercept the entire fan the linear array typically of 500 to 800 separate detectors spreads out. As in previous generations, the gantry rotates the tube and detector array together about the isocenter by making a smooth 360° swing obtaining $500 - 1000$ views during the rotation. With these improvements the image reconstruction time is reduced to $5\,s$ [11].

2.3.4 Fourth Generation scanners

The geometry of fourth generation scanner, introduced in 1978, is presented in Figure 9d [4]. The scanner has a stationary ring of detectors completely circumscribing the patient and only the tube rotates around the patient. Despite this change the image obtaining time is still about $5s$. Since the fourth generation scanner has a ring of detectors instead of an arc, six times more detector elements (up to 5000) are required compared to the third generation scanners. The detectors and their associate electronics are not cheap, and therefore, the seventh generation multi-slice CTs are based on third generation geometry [11].

2.3.5 Fifth Generation scanners

The electron beam computed tomography (EBCT) is referred as the fifth generation scanner [11]. It was built between 1980 and 1984 for cardiac applications, which means the complete set of projections has to be collected within 20 to $50\,ms$. This kind of speed is very challenging for the third and fourth generation scanners since the X-ray tube and the detectors would be exposed to an enormous centripetal force.

Figure 10 shows a presentation of electron beam scanner. The X-ray tube is fixed and the rotation of the source is provided by the sweeping motion of the electron beam [11]. The bottom arc (green dashed line in the figure) represents an anode with multiple target tracks. A high-speed electron beam is focused and deflected by carefully designed coils to sweep along the target ring which is similar to a cathode in an X-ray tube. This whole assembly is sealed in a vacuum. From the target ring fan-shaped X-ray beams are produced and collimated to a set of detectors on the other side of the ring. When using multiple target rings and detector rings an $8\,cm$ long coverage of patient can be obtained for the heart. Due to the lack of mechanically moving parts $50\,ms$ scan times can be achieved.

There are some problems with the fifth generation machines. They are complex, large-sized, the electronics is quite noisy, and it is cost-efficient only with a steady flow of cardiac patients [4]. Today the seventh generation scanners have replaced practically all the fifth generation scanners.

2.3.6 Sixth Generation Scanners

The first generation six machines were introduced in 1989 [11]. Previously there was no movement along the axis of the patient, the z-axis, during projections. The sixth generation scanners combine the movement of the X-

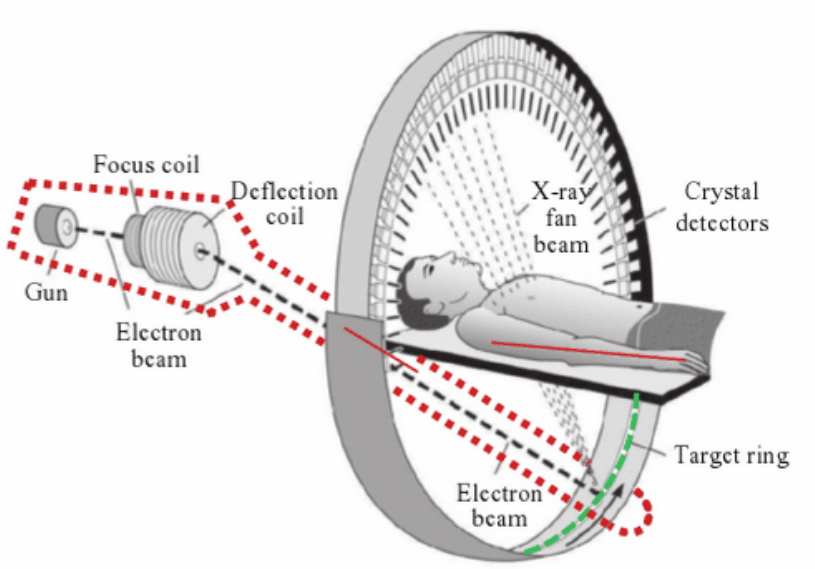


Figure 10: Electron beam computed tomography is referred as the fifth generation scanner. The X-ray tube is fixed and the rotation of the source is provided by the sweeping motion of the electron beam. [4]

ray tube around with a simultaneous smooth displacement of the patient into the gantry opening (Figure 11). The geometry of the detectors can be either the third or fourth generation and the projection system moves in a spiral path around the patient [4]. The sixth generation machines are referred as single-slice spiral CT (SSCT).

Sixth generation scanners differ from the pre-helical design in several ways [4]. Since the table and the gantry move continuously and smoothly throughout data acquisition, the scanning is much faster: the scan times are reduced typically to $1\,s$. Because the tube rotates a cork-screw like path around the table, there are no natural transverse planes so they must be produced artificially [1].

In third and fourth generation scanners the high-voltage electric power to the X-ray tube and the electric signals from the detectors are carried directly by permanent cables which limits operations to a single gantry revolution each way [4]. In sixth generation scanners the problem is solved by the slip ring and conducting brushes enabling helical approach. The connecting brushes rub against the slip rings, which are fixed on the rotating gantry, connecting the gantry to the stationary part of the scanner.

The term detector pitch is used and is defined as table distance traveled in one 360° gantry rotation divided by beam collimation 2.

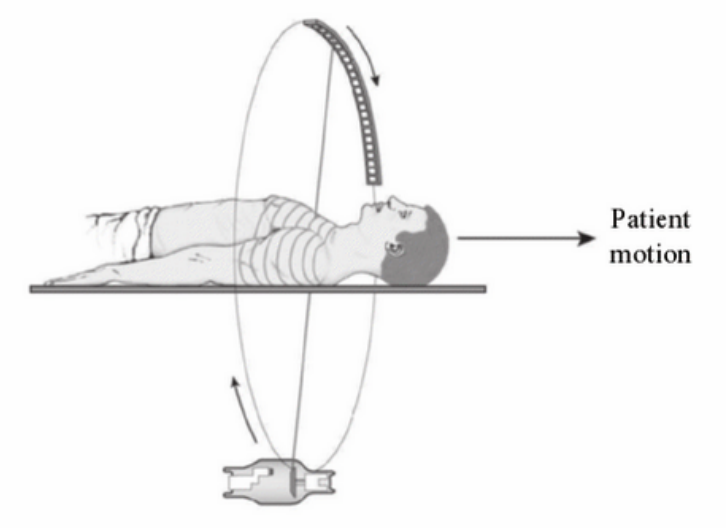


Figure 11: In sixth generation scanners the table moves along the axis of the patient during the projections. [4]

Helical imaging has an adjustable determinant called pitch which describes how much the table moves compared to beam width in 360°rotation [4]. Pitch it is defined as [4]

$$Pitch = \frac{table\ z-travel\ /\ 360^{\circ}\ rotation\ (mm)}{beam\ width\ in\ z-direction\ (mm)} \tag{2.8}$$

When pitch is 1 the table moves one beam width forward while rotating through 360°. If pitch is less than 1 the beam width has some overlap at each view angle. If pitch is more than 1 some view angles are not covered by the beam width at certain table positions. Pitch is demonstrated in Figure 12. Typically pitches range from 0.7 to 2. Higher pitch lowers the radiation dose and scan time, but on the other hand, increases the image noise and resolution.

2.3.7 Seventh Generation Scanners

The seventh generation helical multi-slice scanners were introduced in 1998 [2]. In previous generations there was only one ring or row of detectors, which was exposed by a thin fan-beam of X-rays. In the seventh generation scanner the projection system still moves in a spiral path but the detector array is made of 8 to 34 rows of detectors making it possible to obtain four adjacent slices simultaneously. Thus, the scan times reduce to only $500\,ms$ [1]. This

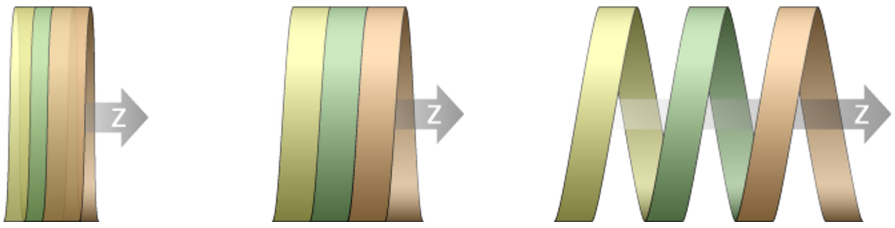


Figure 12: Helical approach was enabled by pitch. When pitch is one the table moves a beam width forward while rotating 360°(center). If pitch is less than one the beam width overlaps at each view angle (left). If pitch is more than one some view angles are not covered (right). [30]

scanner is referred to as a multi-slice spiral CT (MSCT). In MSCT the X-ray beam takes the shape of a cone. The design allows three-dimensional projection techniques to be mastered and thereby paves out the way for the reconstruction techniques operating in three-dimensions.

Between the years of 2001 and 2002 the first proper cone-beam spiral CTs (CBCT) were taken into use [2]. With the cone-beam it is possible to increase the detector array width to 16 or even 320 elements. This allows the simultaneous acquisition of up to 320 adjacent image slices. The cone-beam CT increases the scanning speed and reduces the impact of collimation inaccuracies on the quality of the reconstructed image. The reduction of the collimation losses allows a decrease in the power of the X-ray tube. The spiral motion and the cone-beam together reduce the total scan time. It is also possible to reconstruct many adjacent slices of images during the scan.

2.4 Standard CT Scanner Configuration

Despite many generations of CT scanners the main elements remain the same. The main components are the X-ray tube and high-voltage generator, the X-ray detector and data acquisitions system (DAS), image reconstruction system (IRS), and patient table [11]. Other major elements in the CT scanner are the gantry and slip ring, and collimation and filtration. Everything is controlled by computer.

The most recognizable element of a CT scanner is the 'donut' shaped gantry, where the X-ray tube and detectors are generally placed opposite to each other and they rotate synchronously around the gantry aperture [2]. The continuous rotation in one direction without cable wrap is possible with slip rings, which enable the transmission of power and electrical signals from stationary to the rotating structure. The projections obtained from

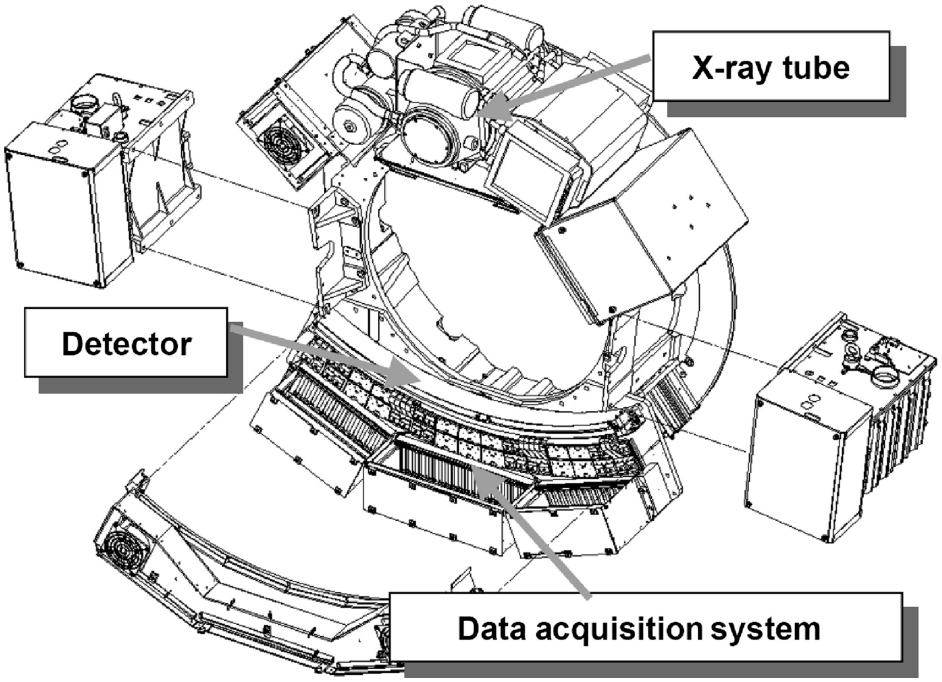


Figure 13: Illustration of a CT scanner. The most recognizable element is the gantry where the X-ray tube and detectors are (generally) placed on opposite sides. [3]

the X-rays are converted from radiation intensities to electrical quantities in the detector array. The image reconstruction system refers to the computer hardware performing reprocessing, image reconstruction, and postprocessing. The patient table allows the patient to be moved easily into position. The table can be moved manually before the scan but moves automatically during it. The patient can be appropriately positioned depending on the body part being examined since the table moves into or out of the gantry along the axis of the patient's body as well as up or down.

2.4.1 X-ray Tube and High-Voltage Generator

Since its invention by Röntgen in 1985, the size and appearance of the X-ray tube have changed significantly but the fundamental principles have remained unchanged [2]. X-ray beam creation requires an evacuated X-ray tube and a high-voltage generator [28]. The evacuated tube holds inside two electrodes: the cathode and the anode. The cathode is heated by driving a constant d.c. current through a heating filament. As a result electrons are emitted and form an electron cloud around the cathode.

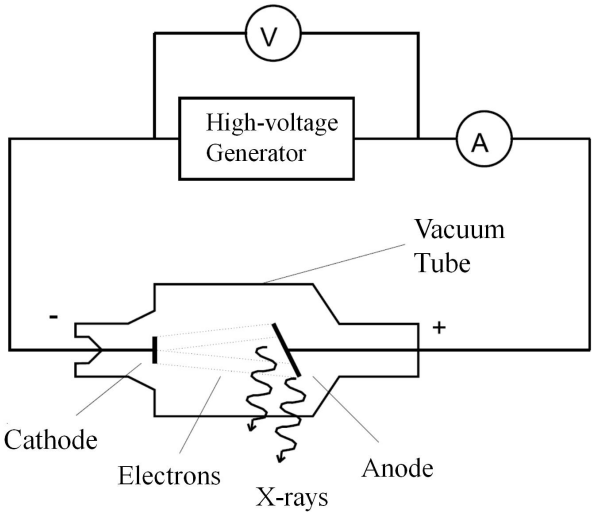


Figure 14: X-ray generation requires a high-voltage generator and an evacuated x-ray tube which holds inside the cathode and the anode. For X-ray generation, a short high-voltage pulse is applied between the heated cathode and the anode. As a result electrons accelerate and hit the anode with a significant speed creating X-rays. Modified from [26]

X-rays are generated by applying a short high-voltage pulse between the cathode and the anode which creates a strong electrical force between the electrodes [28]. As a result electrons from the cathodes electron cloud accelerate and hit the focal spot of the anode with a significant speed. A typical size of the focal spot is in order of $0.6 - 1.2\,mm$ but can be down to $0.5\,mm$.

Three factors can be used to control the X-ray exposure from the generator [28]. The tube potential (kVp, "peak kiloVoltage") affects the energy of the X-rays: the power of the penetration of the X-ray energy and the amount generated during the exposure can be increased by increasing the potential. The total produced X-ray beam energy is directly proportional to the tube current (mA, in milliAmperes) and the exposure time ($mA \times s$). The total energy to the anode from the generator in an exposure is $kV \times mA \times s$. However, only a small fraction of this results to X-rays since most of it is just transferred as heat. CT scanners today generally produce X-rays continuously ranging at $80 - 140\,kVp$ with currents up to $800\,mA$ [1, 4].

2.4.2 X-ray Detector and Data Acquisition System

The primary design considerations for X-ray detectors are detector efficiency, stability of operation, short response time, low dependence of detector re-

sponse on X-ray tube energy, and finally the cost [4]. The detectors can detect X-ray photons in two ways: directly or indirectly.

Most commercial scanners today use indirect detectors, which consist of scintillator crystal together with a photodiode [3]. This detector type is presented in Figure 15. The scintillators convert the coming X-ray photons to visible light which is then detected using conventional photodetectors. The created photons travel in all directions [11]. To direct the photons to the photodiodes, the scintillator is coated with a highly reflective material. However, due to reflection and absorption in the scintillator, only a fraction of the photons reach the photodiodes and create electrical signals.

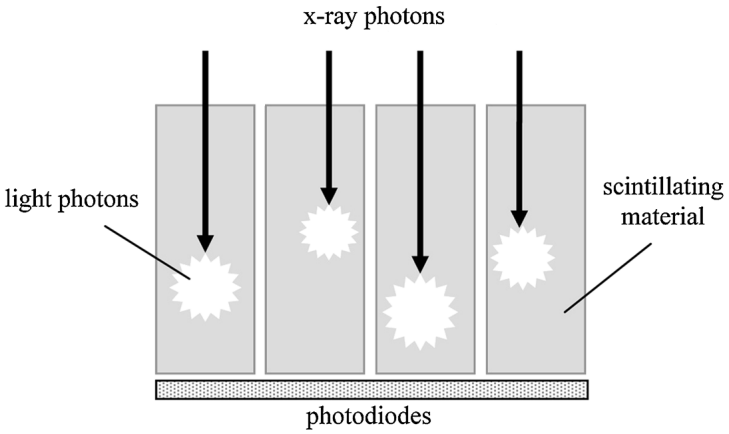


Figure 15: Illustration of a scintillator detector. Scintillators convert the coming X-rays to visible light which is then detected using conventional photodetectors. [11], modified

In direct detection semiconductor detectors convert the X-ray photons directly into electrical signals [11]. The semiconductor detector structure is presented in Figure 16. In this detector an X-ray photoconductor with a bias voltage is applied across the detector structure. When the high-energy photons strike the photoconductor, they create electron-hole pairs. These pairs are then swept by a high electric field to the opposing electrodes creating a charge pulse. The attractiveness of semiconductor detectors arises in the potential performance of X-ray photon counting. However, the detectors do not have currently sufficient count rate capability to handle the X-ray flux required in clinical CT applications.

Each detector channel is connected to the data acquisition system (DAS) [27]. DAS integrates the photocurrent and converts the electric charge signal to voltage using a transimpedance amplifier. DAS is also responsible for the analog to digital conversion with typical sample rates on the order of a couple

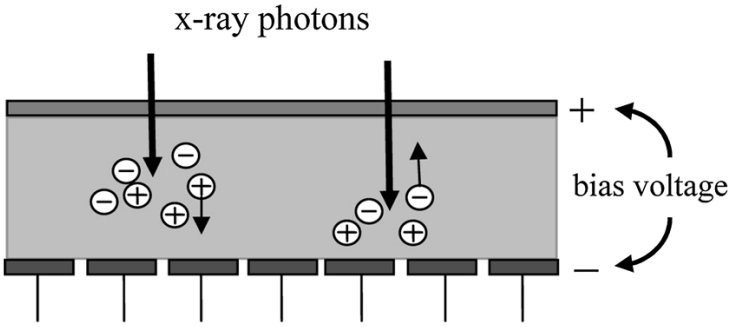


Figure 16: Illustration of a semiconductor detector. X-rays are converted directly to electrical signals. [11], modified

of kHz.

2.4.3 Collimation and Filtration

The purpose of the collimation is to ensure good image quality and to reduce patient dose [11]. Collimation can also be divided in to two categories: prepatient collimation and postpatient collimation.

The X-ray photons emitted from the X-ray tube cover a wide range along the z-axis [11]. By positioning prepatient collimation between the patient and the X-ray source the X-ray flux to the patient is restricted to a narrow beam as shown in Figure 17. When using a single-slice CT the prepatient collimation also defines the slice thickness of the imaging plane. However, this is not the case with multi-slice CT where the slice thickness is defined by the detector aperture. Since prepatient collimation blocks over 90% of the emitted X-rays the efficiency of the tube is quite poor.

Prepatient collimation divides the X-ray beam to two regions: umbra region and penumbra region [11]. In the umbra region the X-ray flux is homogeneous. Respectively, the flux is nonhomogeneous in the penumbra region. With single-slice CT the slice thickness is defined by the full width half maximum (FWHM) and full width at tenth maximum (FWTM) of the entire umbra-penumbra region. FWHM represents the distance between two points on the slice sensitivity profile (SSP, a curve showing the effect of broadening of CT slice thickness along the patient axis in helical CT) whose intensity is 50 % of the peak. Respectively, FWTM represents the distance between two points on the SSP whose intensity is 10 % of the peak. For multi-slice CT these regions have an important role in patient dose. In most commercial scanners only the umbra region is used for image formation and the penumbra region remains unused. The unused X-ray photons therefore

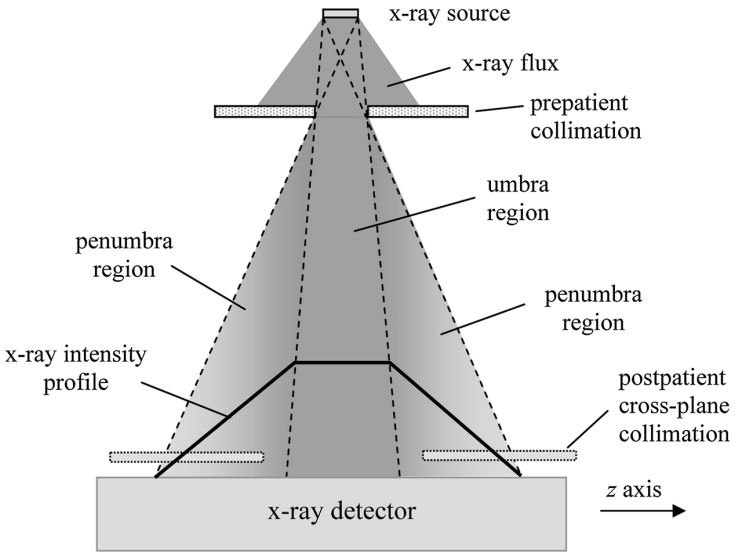


Figure 17: Illustration of prepatient collimation. By placing a collimator between the patient and the X-ray source the X-ray flux to the patient is restricted to a narrow beam. Modified from [11]

play an important role in dose efficiency.

Two types of collimators are typically used for postpatient collimation: one for rejecting scattering and the other for improving resolution [11]. Scattered radiation travels most likely in a direction different from its original path. To prevent it from reaching the detectors attenuating plates in-plane (1D) or cross-plane (2D) are placed in front of the detector (Figure 18). As stated above, the umbra and penumbra regions determine the slice thickness. However, it is difficult to design a prepatient collimation that provides very thin slice profiles. Therefore, additional collimation near the detector surface can be employed to further restrict the X-ray beam to a narrow slice thickness. Similarly the inplane resolution can be improved by blocking partially the surface of detector cells with a comb-style filter.

The X-ray photons emitted from the X-ray tube have also a wide spectrum as seen in Figure 4. However, the low-energy X-rays are primarily absorbed by the patient and contribute little to the detected signals. Therefore, to reduce the patient dose, these soft X-rays have to be removed. This is achieved by improving beam quality with additional X-ray filtering. The most common filters are the flat filter and the bowtie filter. The flat filter is placed between the patient and X-ray source. Hence the X-ray spectrum is uniformly modified across the entire field of view (FOV) by the filter. However, since the cross-section of a patient is more of an oval-shaped a bowtie

filter can be employed to reduce the patient dose even further by modifying the X-ray beam intensity inside the FOV. Both filters are illustrated in Figure 18. [11]

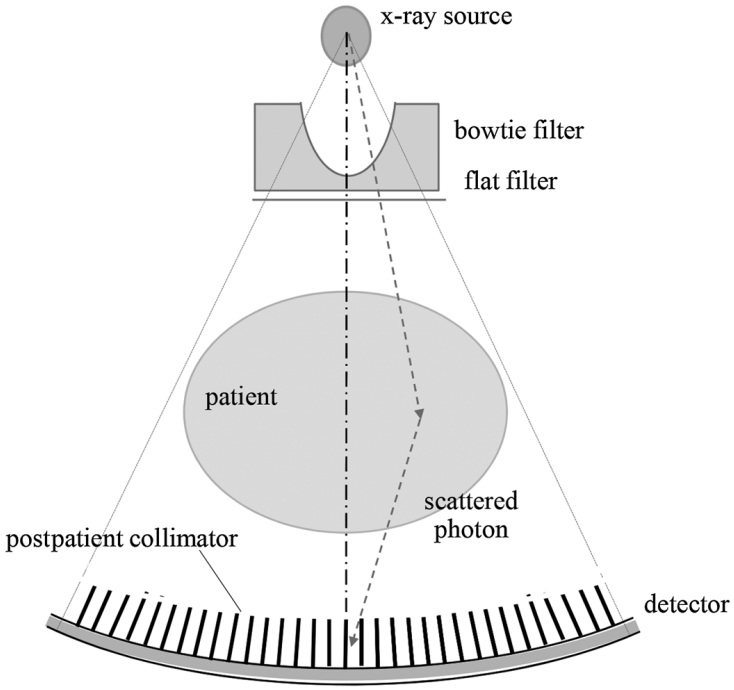


Figure 18: Schematic illustration of the postpatient collimation and X-ray filtration. [11]

2.4.4 The Gantry and Slip Ring

Gantry is an important part of a CT scanner since it houses all the necessary components to produce and detect X-rays to create a CT image [11]. The rotating gantry typically contains the X-ray tube, cooling tank, high-voltage generator, X-ray detector, slip ring, and some other supporting devices. With all these components the weight of the gantry comes to more than $1000\,kg$. Despite the huge weight, the gantry has to maintain angular and position accuracy. Angular accuracy demands the gantry to rotate at highly constant speeds. Position accuracy demands the gantry to be free of significant vibrations in both in-plane and cross-plane.

In addition to gantry, the slip ring is another important component of a CT system [11]. The slip ring supplies power to the rotating gantry, transmits commanding signals, and sends the projection data to the stationary side. The slip ring connection, which can be electrical, optical, or RF. The gantry is

connected to the rest of the scanner with silver or carbon fiber brushes which rub against the slip rings [4]. A CT scanner requires about half a dozen slip rings and each of them has to be electrically insulated from everything else and the brushes must ensure noiseless electrical contact.

2.5 Clinical use

CT is applied for obtaining anatomical images of all parts of the human body [27]. The most common examinations include the following:

- head and neck: brain, maxillofacial structures, inner ear, soft tissues of the neck
- thorax: lungs, chest wall and mediastinum, heart, large vessels
- urogenital track: kidneys, adrenals, urinary bladder
- abdomen: gastrointestinal tract, liver, pancreas, spleen
- musculoskeleton system: bone fractures, soft tissue tumors, muscle tissue

In general, in CT scan the patient is positioned to the patient table and a scanogram or scout view is prescribed [11]. This scan determines the patient's anatomical landmarks and the exact location and range of the CT scan. During scout view the patient table moves while the X-ray tube and the detector remain stationary so it is similar to a conventional X-ray. After the scout scan the gantry rotates to the desired orientation as prescribed by the operator and the actual scan can begin. During the scan the table moves on a constant speed and the high-voltage generator keeps the current and voltage to the X-ray tube at the prescribed level. During the scan the DAS samples the detector outputs at a uniform sampling rate and converts analog signals to digital ones. The sampled data is then sent to the IRS for processing. The data is preprocessed and enhanced before it is sent to the the operator viewing on the display and to the data storage for archiving.

Chapter 3

Radiation Dose

3.1 Terminology

3.1.1 CT Dose Index

The CT dose index (CTDI) is currently the primary metric used to describe the radiation output from a CT scanner [18]. CTDI is defined as the dose absorbed by a standard cylindrical acrylic phantom for one 360° rotation of the X-ray tube [27]

$$CTDI = \frac{1}{n\triangle z}\int_{-\infty}^{\infty} D(z)dz. \tag{3.1}$$

Here n is the number of detector rows, $\triangle z$ the slice thickness, and $D(z)$ the radiation dose profile along the z-axis. The unit is milligrays (mGy).

CTDI is measured by using a pencil ionization chamber placed in a phantom [31]. The two standard phantoms used for CTDI measurements are a head phantom with a diameter of 16 cm and a body phantom with a diameter of 32 cm. The most commonly used ionization chamber for the measurements is 100 mm long and the obtained CTDI is therefore referred as $CTDI_{100}$. The CTDI is generally measured at the center and at four peripheral locations of the phantom.

However, CTDI is not constant and varies across the image plane [27]. At the periphery it is higher than in the center of the FOV. To tackle this problem, a weighted CT dose index $CTDI_w$ is introduced

$$CTDI_w = \frac{1}{3}CTDI_c + \frac{2}{3}CTDI_p, \tag{3.2}$$

where $CTDI_c$ is the CTDI value at the center and $CTDI_p$ at the periphery of the phantom. The relative areas of the periphery and the center are approximated.

CTDI is originally defined for circular scan protocols but with helical scans the pitch affects the absorbed dose [27]. This is taken into account by the volume CT dose index $CTDI_{vol}$

$$CTDI_{vol} = \frac{CTDI_w}{pitch}. \tag{3.3}$$

$CTDI_{vol}$ is mostly affected by the tube voltage and current, and pitch [32]. It is a useful indicator of the radiation output for specific exam protocols [31]. However, it is not a direct measurement of dose and is primarily a measure of machine output.

3.1.2 Dose Length Product

Dose-length product (DLP) is the second important metric for evaluating the patient dose [32]. CTDI is an estimate of average radiation dose only in the irradiated volume [31]. However, risk caused by ionizing radiation is more related to the total amount of the radiation dose deposited in the patient. To better represent the delivered energy in a scan protocol, $CTDI_{vol}$ is integrated along the scan length to calculate DLP [31]:

$$DLP = CTDI_{vol} \cdot l, \tag{3.4}$$

where l (in cm) is the scan length in z-direction. In addition, DLP is affected by the number of acquisitions (irradiation events). The drawback of DLP is that it does not take into account the radiosensitivity, relative susceptibility to harmful effects of ionizing radiation, of the irradiated tissues, and therefore, is not an appropriate risk indicator.

3.1.3 Effective Dose

Effective dose (ED) reflects the stochastic risk from an exposure to ionizing radiation and is typically expressed in millisieverts (mSv) [18]. ED takes into account the radiosensitivity of the organs and the equivalent doses to all exposed organs [33].

ED is obtained by multiplying DLP by a tissue and organ specific constant κ [4]:

$$ED = DLP \cdot \kappa. \tag{3.5}$$

The conversion factor κ is obtained from Monte Carlo calculations based on generic patient, thus, ED is not an appropriate risk indicator for an individual patient.

ED is practical for comparing different radiological exposures on common scales [32]. With ED it is possible to compare the medical examination dose to other forms of radiation, for example, cosmic rays or occupational radiation exposure.

3.1.4 Size-specific Dose Estimate

Size-specific dose estimate (SSDE) is a patient dose estimate [34]. It takes the patient size into account with a correction factor which is determined using linear dimensions measured from the patient.

For calculating the SSDE, the effective diameter of the patient needs to be determined [34]. Assuming the patient has a circular cross section, the effective diameter presents the diameter of the patient at a given location along the z-axis of the patient. However, in real life not all body parts are circular, thus, a circle with the same area as the cross section of a patient is used instead. Assuming the patient is elliptical, the effective diameter is [34]

$$D = \sqrt{A \times L}, \tag{3.6}$$

where A is the anterior-posterior dimension, and L is the lateral dimension (side-to-side) of the scanned body part.

The effective diameter determines the correction factor for a reference phantom which can be either $16\,cm$ or $32\,cm$ diameter $CTDI_{vol}$ [34]. The correction factors corresponding these are f_{size}^{16X} and f_{size}^{32X}. The X refers to the specific measure of size used, where $X = S$ for the sum of AP and lateral dimension, L for lateral, A for AP, and D for effective diameter. The correction factors are calculated by American Association of Physicist in Medicine (AAPM).

SSDE is calculated by multiplying $CTDI_{vol}$ with a correction factor by [34]:

$$SSDE = {f_{size}}^{32X} \times {CTDI_{vol}}^{32}, \tag{3.7}$$

for the $32cm$ diameter reference phantom, and [34]

$$SSDE = {f_{size}}^{16X} \times {CTDI_{vol}}^{16}, \tag{3.8}$$

for the $16cm$ diameter reference phantom.

The dose at the center of a certain CT scan can be estimated with SSDE [35]. However, it does not take into account dose variations based on variations in scan length and is affected by the CT radiographs, patient contour along the x-axis, and patient positioning in the gantry. Finally, SSDE can not be used for organ dose estimation, and therefore, can not be used for effective dose estimation.

3.1.5 Absorbed Dose

Absorbed dose is most commonly referred as the patient dose. It describes the amount of ionizing radiation deposited in tissue (in mGy) giving the amount of energy (J) deposited per unit mass (kg). For example, absorbed dose differs between two patients of different size scanned with the same $CTDI_{vol}$. The absorbed dose in the smaller patient is higher because with the larger patient the same radiation dose is distributed over a larger mass. Absorbed dose can either be calculated or estimated from SSDE and is dependent on DLP, patient size, and scanned organs.

3.2 Radiation Dose and Quality Factors

Radiation dose and image quality are affected by several patient factors and CT parameters [36].These factors need to be identified and used in creation of CT protocols for different body regions and clinical indications.

Patient size always affects the radiation dose [36]. For same clinical indication and body region, a larger patient usually requires higher radiation exposure to obtain the same image quality as with smaller patients. Therefore, especially small children should be scanned with much lower radiation doses compared to adults. Similarly, tissues with smaller X-ray attenuation should be scanned with lower radiation doses compared to the ones with higher attenuation.

Clinical indication for CT is itself an important factor because changes in radiation dose or image quality affect the conspicuity of different findings [36]. For higher tissue contrast, CT examinations can be performed with lower radiation doses, because image noise does not harm the detection of these high-contrast structures. In addition, the lesion conspicuity is also affected by the lesion size, location, contrast enhancement, and cross-sectional area of the patient. Due to these factors it is important to pay attention to CT protocols: they should be designed to take into account the scanned body region, cross-sectional area of the patient, and examination indication.

The radiation dose and image quality are also affected by many scanning factors, for example, tube current, tube voltage, gantry rotation time, detector configuration, and pitch [36]. Some of these factors can be automated and others need to be manually adjusted. The ability to adjust the factors depends on the CT system in use and different vendors have their own proprietary nomenclature for similar scanning factors. This can cause confusion if scanners from different vendors are used in same institution.

3.3 Image Quality

Image quality is closely related to radiation dose and is affected by the same factors [36]. Image quality depends on four basic factors: spatial resolution, noise, contrast, and artifacts [31]. Depending on the diagnostic task, these factors determine the sensitivity and the visibility of details. Sensitivity refers to the ability to perceive low-contrast structures.

3.3.1 Spatial Resolution

Spatial resolution is the ability to differentiate small, closely spaced objects in an image [31]. Resolution depends on multiple factors including: focal spot size and penumbra, width and the length of detector element, binning, FOV, pitch, nature of interpolation, reconstruction kernel or convolution filter, scatter or other noise, and patient movement [4].

CT resolution is mostly restricted by sampling - the size and spacing of samples used to reconstruct an image [31]. This is illustrated in Figure 19. The sample can not be resolved if the sample size (aperture) is too large (Figure 19a), or the samples are too far apart from each other (Figure 19b). When the sample spacing and aperture size are adequate the pattern can be resolved (Figure 19c). However, if the position of samples relative to the pattern is not right, the effective resolution may be lower than expected (Figure 19d). Sometimes also aliasing can occur. In aliasing the pattern seems to be resolved but with incorrect number of details (Figure 19e).

In addition to sampling, focal spot size can to some extent affect the resolution and patient motion can cause blurring [31]. Spatial resolution is always affected by the reconstruction kernel or convolution filter which can either suppress high frequencies for noise and aliasing reduction or enhance the high frequencies for the sharpest images. The displayed resolution is linked with the achievable pixel size [4]. The dimension of a pixel should be about a half or less of the system's required in-plane resolution. If the pixel size exceeds the spatial resolution, it becomes the bottleneck and determines the spatial resolution.

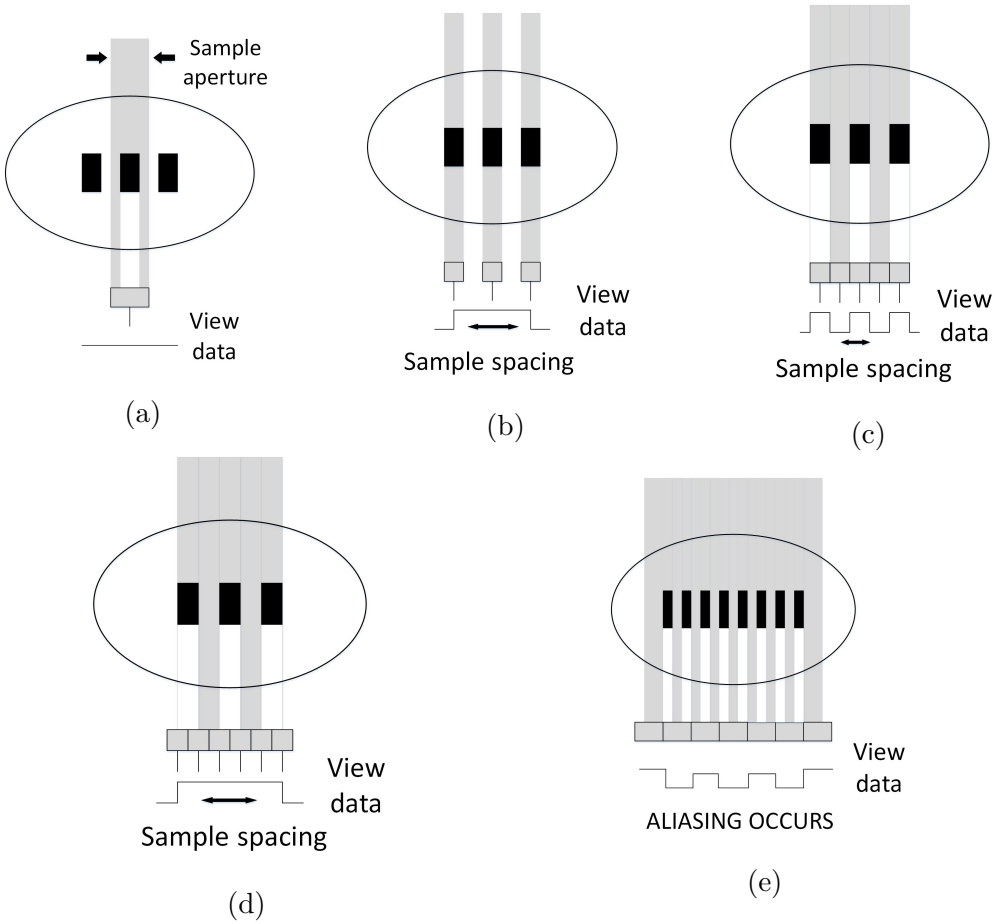


Figure 19: Resolution of CT system is limited by sampling-size and spacing of samples used for image formation. The pattern is unresolved if (19a) sample size (aperture) is too large, or (19b) samples are too far apart. The pattern is resolved only when(19c) aperture size and sample spacing are adequate. (19d) Effective resolution can be lower if sample positioning relative to patter is poor. (19e) In aliasing the pattern seems to be resolved but with incorrect number of details.

3.3.2 Noise

The three distinguished types of noise in CT are quantum or statistical noise, electronic noise, and round-off or quantization noise [27]. Due the nature of X-rays, the main contributor is quantum noise. A photon detection is essentially a Poisson process where the variance is equal to the mean. Due to this the noise amplitude is proportional to the square root of the signal amplitude. As signal-to-noise-ratio (SNR) behaves as the square root of the

signal amplitude, reducing dose decreases also the SNR.

The appearance of noise can be affected by the reconstruction algorithm [31]. The visual impact of noise can be reduced with smooth filters and enhanced with sharp filters. Smooth filters are preferred with soft tissue where the noise is more interfering than blur. With structures containing edges and small details blur is typically more interfering than noise and sharp filters are preferred.

3.3.3 Contrast

Image contrast is dependent on the subject contrast and the display contrast [27]. The subject contrast is primarily dependent on the attenuation properties of the patient. However, physical factors such as X-ray tube spectrum, extent of beam-hardening and scatter, and detection nonlinearities have an effect on on the contrast. The display contrast can be modulated by the gray level transformation after the image formation, leaving noise as the main limitation on low-contrast detail perception.

3.3.4 Artifacts

Artifacts are artificial structures in the image, which deviate from reality [31]. The three main categories for artifacts are shading artifacts, ring artifacts, and streak artifacts.

Shading artifacts often appear near objects with high contrast [31]. The most common type of shading artifacts is beam-hardening which is caused by the polychromatic X-ray beam spectrum and the energy dependent attenuation coefficients [11]. As the beam passes trough an object, lower energy photons are absorbed more rapidly than the higher-energy photons and as a result the mean energy of the beam increases ("hardens") [37]. Beam-hardening appears as nonuniformities in the CT numbers, for example, CT numbers of an uniform phantom are lower at the center than at the periphery. Normally the uniformities are quite small ($< 5\ HU$), thus, they are not apparent. However, when a scan is is passing through a thick region (e.g. bone) larger beam-hardening can occur. This causes mostly dark hypointensity regions in the image. Shading artifacts can also be caused by scattering but this is uncommon in most scanners today. Shading artifacts often mimic pathology and can lead to misdiagnosis, and therefore, should be handled with great caution.

Ring artifacts are associated mainly with the scanners based on third generation and, as the name applies, appear as rings or arcs on the original image structure [11]. Full rings are dissimilar to human anatomy, thus, they

can be recognized as artifacts. However, partial rings can mimic certain pathologies and may wind up in the image. Ring artifacts are caused by errors, imbalances, calculation drifts, or other measurement inaccuracies in an element of a detector array relative to the adjacents [31]. Ring artifacts can usually be removed from the image since they are easily recognizable by ring-correction algorithms. However, small rings and arcs may pass the algorithm and end up in the image.

Streak artifacts can occur in all scanner types and they often appear across the image as intense straight lines that can be either dark or bright [11]. Streak artifacts can arise for many reasons but most of them are due to bad or inconsistent detector measurement [31]. Inconsistencies include factors like motion, partial-volume effects, metal, insufficient X-ray intensity, and malfunctions. Inconsistencies create streak artifacts because of the nature of the back-projection. They can be recognized and corrected with algorithms in some cases. Streak artifacts may also be minimized or avoided through appropriate scanning techniques or using thinner slices.

Chapter 4

Quality Assurance

4.1 Principles

The best way to fulfill the requirements of the Radiation Act of Finland is to have a comprehensive quality system in use [38]. Quality system covers the organization structures, procedures, and resources of quality management. Quality management is described in quality documents, which form a coherent and up to date ensemble that in practice means a quality manual or equivalent. Quality assurance (QA) is part of quality management and technical quality control is one of the most essential elements in QA. Other important aspects of QA are patient dose, reference levels, image quality evaluation, re-analysis of images, self-evaluation, and clinical audition. The main structure and the most important aspects of quality management are presented in Figure 20.

QA of radiological equipments means a constant surveillance of the working order and functional qualities of the equipment [38]. This surveillance is present at every step of the equipment's life cycle and it is the key part of QA for X-ray examinations. The goal is to ensure the desired clinical benefits are achieved and the patient's X-ray dose will not exceed the necessary levels required for diagnosis.

Technical QA sets the basic requirements for carrying out optimizing principles in radiation protection [38]. Technical QA prevents bad image quality before appearing in patient X-ray images and it takes into account the safety of the personnel using the X-ray systems. Technical QA is a fixed part of running the equipment. For this reason the resources required for it must be seen as a part of the operational expenses.

Today digital techniques are widely used and they come with some important aspects that need special attention [38]. The distribution and archiving

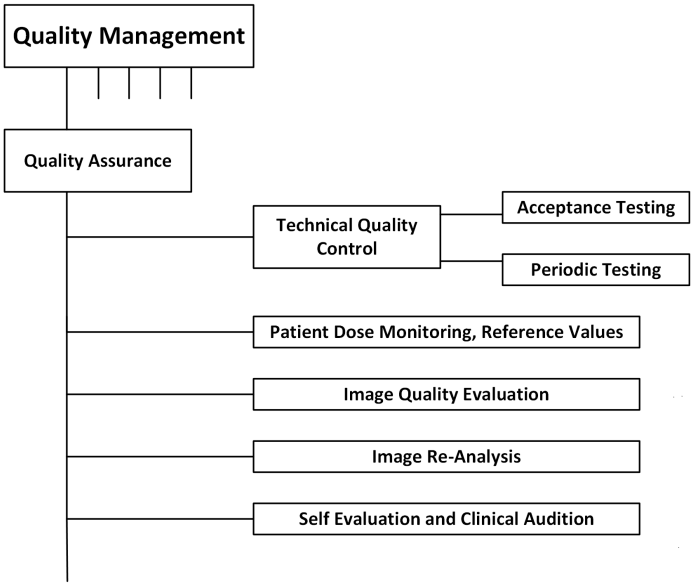


Figure 20: Main structure of quality management.

of digital X-ray images and patient histories have become electronic. Therefore, it is important to pay attention to standardizing the presentation of digital X-ray images. In practice this means standardizing the image displays, and processing and storing of the images. This way the images in the archive would be diagnostic with color scale without additional image processing in the workstation. The standardization requires optimizing the imaging instructions and image processing together with the radiologist and the radiographer.

4.2 Phases

The life cycle of a CT system has many phases starting from the purchasing process, which results in the installation and acceptance testing of the equipment [38]. Purchasing process is followed by commissioning testing and defining the reference values for the equipment. After this the Radiation and Nuclear Safety Authority (Säteilyturvakeskus, STUK) tests the performance of the equipment and finally the clinical use can begin. During clinical use STUK continues to supervise the performance of the equipment, and maintenance and quality control are performed.

QA is performed several times during the life cycle of the equipment. The QA tests can be divided into the acceptance tests and the periodic

tests. According to the Decree of the Ministry of Social Affairs and Health (423/2000, § 32) the functions of a radiological equipment must be tested

1. before the introduction of a new equipment (acceptance test);
2. at specific intervals according to device-specific instructions (periodic testing);
3. after major repairs or maintenance; and
4. when there is a reason to suspect a malfunction or a change in operation of an item of equipment. [24]

4.2.1 Acceptance Tests

When a new equipment is installed, the supplier and the mechanic are responsible for ensuring the equipment works properly and safely after the installation [38]. In addition to the radiation safety, this includes electrical and mechanical safety.

The operator is responsible for the safe implementation and use of the equipment, and either performs an acceptance test by itself or orders it from other party [38]. The aim of the test is to ensure the equipment and all supplies included are undamaged and delivered according to the order, and that the important documents, especially user instructions and guides, have been handed over to the operator. The equipment needs to meet all the performance and safety characteristics according to the delivery contract, manufacturer, and legislation [39]. Usually the tests and measurements required for the clinical use are seen as a part of the acceptance test.

During the acceptance tests it is important to determine the reference values for constancy testing [38]. This means all the tests that are later performed as a part of the QA are performed for the first time as a part of the acceptance test. By comparing the original test results and the ones obtained later during the use it is possible to detect problems related to the functions of the equipment.

The acceptance test can be performed by a representative from the operator, supplier, or a third party [38]. If the operator performs the test itself, it is recommended that the final users take part in it. Generally it is appropriate that the operator and the supplier do the testing together. If the operator, however, does not participate the testing, they are required to ensure adequate monitoring from their side and name a responsible person for it [39].

4.2.2 Periodic Tests

Periodic testing during clinical use includes safety and operation tests [38]. Operation tests ensure the system meets up with its operating targets, especially on patient dose and image quality related factors. Operation tests are usually constancy tests observing if the test results keep within operation boundaries (section 4.3). It is important to perform the test in a consistent way with the acceptance tests to guarantee that the test results are comparable. For some tests, however, the reference values can be based on acceptance limit values instead of the ones obtained during acceptance tests. The safety tests verify the condition of radiation detectors, warning lights, and radiation protections, and the safety of mechanical functions (i.e. emergency switches).

Periodic testing is performed according to the official QA program [38]. Unless there is reason to suspect malfunction of the equipment, there is no need to repeat all the tests included in the acceptance test during periodic testing. However, if signs of malfunction appear, proper QA tests need to be performed.

Official QA program must include documentation about

- executable tests and measurements and their meaning
- procedures for these tests and measurements
- used devices and tools
- periods for the tests and measurements
- acceptable operating values for the tests and measurements
- actions if the operating values are exceeded [38].

In addition to these, the performing party and a person in charge for every test and measurement need to be assigned. However, it is not obligatory to mention the names of the performing parties. It is important to describe the procedures for the tests and measurements thoroughly so they can be performed as desired.

The results from the QA tests need to be recorded [38]. These records must include the names of the tests or measurements performed, the name of the performer, and the results. Also, if the acceptable operating values have been exceeded, the following actions need to be recorded. These records are evaluated regularly and the QA program will be modified on demand.

4.3 Operation Boundaries

QA program determines the operation boundaries the test values need to meet [38]. If the limits are exceeded, the user must take actions to improve the functions and performance of the equipment. The operation boundaries are divided into acceptability and repair boundaries.

If the operating value is exceeded in one QA test, the first step is to make sure whether the test is performed correctly and whether the used devices and tools are working properly [38]. After this the measurement should be repeated. If the operating value still exceeds the limits, necessary actions need to be started.

4.3.1 Acceptability Boundaries

Acceptability boundaries are the minimum requirements for the equipment given by the authorities [38]. If the acceptability boundaries are not met while testing, broader actions need to be taken or, in some cases, remove the equipment from use. Acceptability boundary violations may focus only to a specific part of the equipment. In these cases it is enough to remove only that part of the equipment from use. In X-ray examinations the required equipment performance varies and may also depend on imaging indication, which makes it impossible to give the acceptability boundaries for all X-ray features. The acceptability boundaries are given by STUK with a separate decree [40].

4.3.2 Repair Boundaries

Repair boundaries are normally stricter than acceptability boundaries [38]. These boundaries are usually based on the acceptance tests or on the performance level the system can achieve in normal working conditions. If the repair boundaries are exceeded, proper repair actions need to be started. The extent and the urgency of the repairs vary based on how much the normal usage and safety of the system are affected. It is necessary to evaluate whether the use of the system needs to be modified or stopped if the repairs can not be done immediately. Also, some other QA tests may be needed to find out the cause for system performance drop. Finally, all the actions taken need to be reported with other QA test results. The repair boundaries can be set by the operator but they can not exceed the acceptability boundaries [39].

4.4 Measurement Uncertainties

The results of a measurement depend on the measurement procedures, the measuring system, the measurement conditions and environment, and generally on the person performing the measurement [38]. It is important to familiarize and define the measurement uncertainties. Only then it is possible to reliably evaluate the performance acceptability or operation boundary violations based on the measurement results.

Uncertainties in radiation measurements can arise from the energy and dose speed dependency of the meter, repeatability, and the response [38]. Additionally, the measurement distance, radiation quality (the voltage and filtration of the X-ray tube), and the radiation field size can cause measurement uncertainties. Typically these sources of uncertainties will together cause at least an error of 10%. To minimize the uncertainties the geometry of the measurements should be described specifically, preferably with pictures, in the QA instructions.

In general, visual image quality evaluation is based on a test phantom which has a series of targets that either get smaller or deteriorate [38]. During the test, an observer defines the weakest distinguishable target on the phantom. Even though this measurement may seem simple it is actually demanding and obtaining an exact result is difficult. The test requires the evaluation criteria for the test target to be kept the same from measurement to measurement and the same target visibility criteria to be used among test performers. Since obtaining this is demanding, the results vary between measurements and performers, and therefore, only large image quality changes can be detected reliably. The test reliability can be enhanced to some point by using multiple observers and by evaluating the mean of their results. The most reliable and easiest way, however, is to compare the previous and current test images side by side.

4.5 Quality Assurance

STUK has gathered a manual containing the recommended QA tests for CT scanners [38]. The manual explains the meaning and intervals of the tests, and gives guidance on how to perform them. STUK instructs to primarily perform the tests recommended by the manufacturer according to manufacturers' own procedures. If the manufacturer has not given any instructions, it is possible to follow the examples given by STUK. It is also possible to use other methods outside the manual. Primarily the test intervals given by STUK should be followed. However, the test should be performed at least

as often the manufacturer instructs and also when a need occurs. The manual also offers repair boundaries. The recommended tests are each described separately but in practise it is possible to combine multiple test and use same test objects and images. These allow several features to be tested at the same time with a single test. Also, some computer programs exist that enable analysis of the test results automatically and fast.

The manual divides the tests into two: the user tests and the technical tests [38]. From these, the technical tests are generally more comprehensive than the user tests. The tests verifying the core features affecting the system functions need to be included to both tests.

On a regular basis, STUK releases updated lists of acceptability boundaries, which typically concern the accuracy of equipments adjustments and functionality [40]. It is to be noted that these boundaries are not boundary values for optimal efficiency. Therefore, STUK recommends applying stricter boundaries when acquiring new equipments, performing acceptance tests, and monitoring the quality of the equipment. These boundaries can rely i.e. on equipment specifications or tolerance values for efficiency presented in equipment standards.

4.5.1 Quality Assurance Tests

The QA tests include user and technical tests [38]. The users should verify the functions of the equipment every day. On a weekly basis the users should also check the CT value, the noise, and image distortions of an CT image.

The technical tests are more comprehensive, which means there are more tests to be performed. Every six months the slice positioning range accuracy, CT values, uniformity and geometric accuracy of a CT image, accuracy of the patient table position, and high contrast resolution need to be verified [38]. The verification of slice thickness needs to be done every six months in case of axial imaging, and during acceptance test in case of helical imaging. Contrast threshold or low contrast resolution and width of dose profile have to be checked during acceptance test. Finally, the dose display correctness needs to be verified every six months and once a year for all collimations.

Some tests recommended for X-ray tubes, generators, image frames, and image displays also apply for CT equipments [38]. Users must verify once a week the functions of the imaging frame and the functionality of safety devices. Also, on a weekly basis the image display should be checked by evaluating the operating environment and the display's functions with a test image.

Technical tests also include the tests concerning image display: the operating environment check and general visual evaluation should be done once

a year [38]. The filtration of the radiation beam should be tested during the acceptance test or after changing the X-ray tube. If the filter is changed automatically, the check should be done yearly. Finally, also the smoothness of the luminosity in the image need to be checked on a yearly basis.

The QA tests concerning CT equiments are listed in Appendix A. The user tests are presented in Table 10 and technical tests in Table 11. Table 12 gives the user tests and Table 13 the technical tests that apply for CT from the tests recommended for X-ray tubes, generators, image frames, and image displays.

4.5.2 Acceptability boundaries

The decree (423/2000, § 30) by the Ministry of Social Affairs and Health states about general requirements and restrictions that: procedures involving exposure to radiation shall be performed using equipment suited for the said purpose [23, 24]. The requirements and acceptability criteria from the radiation safety perspective are confirmed by STUK. The most recent acceptability criteria are stated in STUKs decree 11/3020/2013 [40].

All CT equipments must have a display showing the radiation exposure of the patient [40]. The value may be DLP or $CTDI_{vol}$. The display value can not differ more than 25 % from the actual value. The equipment must also have a notation of the total filtration. The filtration needs to correspond to at least the value of $2,5\,mmAl$. The demands are also met if the half value length (HVL) of the primary radiation is according to the Table 2. HVL is defined as the the thickness of material required to reduce the intensity of an X-ray to one half of its initial value [41].

Table 2: The lowest half value length (HVL) acceptable for primary radiation [40].

X-ray tube voltage (kV)	**Half value length** ($mmAl$)
50	1,8
60	2,2
70	2,5
80	2,9
90	3,2
100	3,6
110	3,9
120	4,3
130	4,7
140	5,0
150	5,4

The X-ray tube voltage cannot change more than 10 % from the set or indicated value [40]. In addition, when adjusting the voltage from one value to another, the actual voltage change has to be between minimum of $0,5$ and maximum of $1,5$ times the difference of the set voltages. The quantity of electricity or the product of X-ray tube current and imaging time cannot deviate more than $20\,\% + 0,2\,mAs$ from the set value. The X-ray tube current can differ at most 20% and the imaging time (rotation time) at most $20\,\% + 1m\,s$ of the set value.

The radiation output of the X-ray tube has to be monitored [40]. When imaging with fixed imaging values corresponding to clinical use of the equipment or with manually set values, the standard deviation of the measured doses in radiation beam are not allowed to exceed 10%. The standard deviation is calculated from at least five repetitive measurements. With manually set values, the measured doses in radiation beam need to be proportional to the fixed quantity of electricity as [40]

$$\frac{|\frac{K_1}{Q_1} - \frac{K_2}{Q_2}|}{(\frac{K_1}{Q_1} + \frac{K_2}{Q_2})} \leq 0,1, \tag{4.1}$$

where K_1 is the dose corresponding quantity of electricity Q_1, K_2 is the dose corresponding quantity of electricity Q_2, and $Q_1 < Q_2 < 2 \times Q_1$.

The movement of the patient table must follow adequate accuracy [40]. The accuracy is verified by moving the table $30\,cm$, measuring the actual

movement of the table, and comparing these values against each other. The difference between the movement displayed by the scanner and the measured one is not allowed to exceed $3\,mm$. Additionally, the addressed and the actual starting points of the scan cannot differ more than $3\,mm$ from each other.

The image display functions cannot limit the quality of the displayed image in a way it affects the reliability of the diagnose [40]. It is to be noted that the performance of the display is affected by the illumination of the working environment. Therefore, the illumination should not be too bright since it weakens the contrast. In addition, disturbing reflections from light sources cannot occur on a dark display.

The image quality must fulfill the clinical demands of an X-ray examination and the clinical images cannot have any traces of earlier images [40]. An image taken from a homogeneous object is not allowed to contain image distortions that could harm the diagnosis of the patient.

4.6 Radiation Dose Monitoring

Currently several CT vendors and models exist making it hard to keep in track with all the different protocols and radiation doses used [32]. For CT dose monitoring, multiple commercial dose-tracking software packages are available [36]. With these softwares the CT dose metrics from the examinations are compiled in a searchable and analyzable database. Some of the softwares also allow the user to set alerts in case the dose falls outside the expected range for a particular CT protocol. Examples of dose-tracking softwares are listed in Table 3. If no dose-tracking software is used, the radiation doses of the most frequently used examinations must be monitored in another way.

Table 3: Some available commercial dose-tracking softwares. More information about the softwares can be found from their websites.

Vendor	**Dose-tracking software**
Siemens	Teamplay
GE Healthcare	DoseWatch
Philips	DoseWise
Toshiba	DoseRite
Sectra	DoseTrack
Bayer	Radiometrics

Chapter 5

Methods

In this thesis the QA and patient dose monitoring methods in CT were examined by a survey which was sent to selected hospitals in Finland. In addition, visits to the same hospitals were done to monitor the QA tests on the spot. The hospitals included to this thesis were: The Hospital District of Helsinki and Uusimaa (Helsingin ja Uudenmaan sairaanhoitopiiri, HUS), Turku University Hospital (Turun yliopistollinen keskussairaala, TYKS), Päijät-Häme Central Hospital (Päijät-Hämeen keskussairaala, PHKS), Oulu University Hospital (Oulun yliopistollinen keskussairaala), Kuopio University Hospital (Kuopion yliopistollinen sairaala, KYS), and Seinäjoki Central Hospital (Seinäjoen keskussairaala, SJKS). Even though The Hospital District of Helsinki and Uusimaa consists of several hospitals, the QA and patient dose monitoring methods are centralized, and therefore, it was handled as one unit and is referred as one hospital in this thesis. The contact persons in the hospitals were hospital physicists.

Hospitals were first contacted by presenting the thesis subject and inquiring if they can and want to participate in the survey. If they were willing to participate, a visit including QA tests was arranged. After the visit the hospitals were given the survey consisting of questions regarding the QA test and patient monitoring methods. The survey sent to the hospitals is presented in Appendix B.

The survey consists of two main parts: QA and patient dose monitoring. QA is further divided into two parts. The first part includes questions regarding the QA tests and their test setup. The goal is to get a comprehensive understanding of the performed tests: What, how, by whom, how often, and why. The questions regarding this part are presented in Table 4. The second part consists of supporting questions considering the QA tests. The goals are to find out if different methods and practices are used among different parties, and if the amount of testing is adequate. The questions regarding

this second part of QA are presented in Table 5.

In the part concerning patient dose monitoring methods the goal is to find out how the hospitals follow patient dose levels, who does it, and how the acceptable levels for the dose have been determined. The questions regarding patient dose monitoring are presented in Table 6.

Table 4: Questions concerning the quality assurance test performed by hospitals.

Question	**Extension**
What quality assurance test?	The name of the test
How is the test performed?	Please describe briefly the test setup
Who performs the test?	Hospital physicist / radiographer / other
How often is the test performed?	Does the cycle differ from instructions?
Why is the test performed?	Does STUK / vendor / hospital / other demand it to be done?

Table 5: Questions concerning the quality assurance test performed by hospitals.

Question	**Extension**
Have you noticed differences in quality assurance test methods between different parties?	Assuming more than one party performs a test with the same goal. What parties? What kind of difference?
Do you think the amount of tests used for quality assurance is adequate?	Why?
What kind of tests could be added in addition to the current ones?	Why should they be added?
Have you encountered any situations concerning image quality problems where the problems haven't occurred in the tests but appear when imaging a patient?	What kind of situations?

Continued on next page

Table 5 – continued from previous page

Question	Extension
What kind of solutions would you need to be able to monitor the patient dose levels better?	In addition to the current ones in use. Why these solutions?

Table 6: Questions concerning the patient dose monitoring in hospitals

Question	Extension
How do you monitor patient dose levels?	-
What applications and methods do you have in use for patient dose monitoring?	-
Who is in charge of the monitoring?	-
Which examination types do you have under surveillance?	The dose varies between different examinations. How do you differentiate different examinations?
Do you compare the patient dose levels among different equipments?	Some equipments may be used only for specific examinations, which make comparing difficult.
How do you compare the patient dose levels between equipments?	-
How has the acceptable dose level been determined in your hospital?	Does your level differentiate from STUKs recommendations? Do you have a lot of specific examinations that could affect the acceptable level?
How do you notice if the dose level has been exceeded or there are abnormalities?	-
What kind of solutions would you need to be able to monitor the patient dose levels better?	In addition to the current ones in use. Why these solutions?

Chapter 6

Results

The results from the survey are found in Tables 7, 8, and 9. Answers were received from four hospitals: HUS, PHKS, OYS, and TYKS. Table 7 includes the answers for part 1 on QA tests and Table 8 the answers for part 2. The answers concerning patient dose monitoring are presented in Table 9.

As seen from the Table 7, the hospital QA tests performed in hospitals include both acceptance and periodic testing. Acceptance tests are performed in HUS and OYS. HUS ensures image quality with Catphan phantom and dose production with $CTDI$ dose measurements. In OYS the X-ray beam width is checked using Gafchromic films. There are more periodic testing ensuring the system operations, for example, a daily check up including calibration, and a weekly evaluation of the CT-values, image noise, and artifacts. In OYS the technical image quality test, patient table movement, and CT dose measurements are performed biannually. HUS performs the dose measurements annually. The daily and weekly test are performed by the radiographers, and other tests by the hospital physicists or engineers. In addition to these, vendor engineers perform technical tests 2 to 4 times a year. Tests are mainly performed on the demand of legislation, STUK, or vendor, but also tests demanded by the hospital are included.

The second part of the survey focused on overall picture of the QA testing and the results are gathered to Table 8. Based on the answers the QA test setups appear to be consistent, though, some minor differences on methods between vendors appear. Hospitals are mainly satisfied on the amount of QA testing performed currently. However, OYS hopes for tools to objectively measure diagnostic image quality. Also tests regarding the X-ray tube modulation and status are suggested by HUS and TYKS. It rises up that some problems concerning image quality have been encountered despite the QA testing. In HUS certain low contrast artifacts have been noticed and OYS has also encountered severe ring and hourglass shaped artifacts on some

occasions.

The answers regarding patient radiation dose monitoring are gathered to Table 9. Currently HUS monitors radiation doses with manual collection, but is switching to dose collection software in 2018. TYKS and PHKS use DoseWatch or RIS. OYS normally does monitoring with $CTDI_{vol}$ and DLP values, and in abnormal situations with effective dose. All hospitals have some examination types under surveillance, although, some variation appears. Patient dose levels between different equipments are also compared in every hospital. In HUS the comparison is done by using distributions, quartiles, means, and scatters, whereas OYS uses mean $CTDI_{vol}$ and DLP values of a small sample size. TYKS and PHKS can do the comparison via DoseWatch. Considering acceptable patient radiation dose levels, both the values determined by STUK and the hospitals values by their own are used. HUS uses only their own levels based on previous results. If the levels are exceeded, TYKS and PHKS get automatic alert from DoseWatch. At HUS the levels are monitored with means and quartiles, and OYS relies at the moment on the reports from the radiologists. All hospitals have a medical physicist in charge of the patient radiation dose monitoring, except at HUS, where radiation supervisors also take part.

For future improvements TYKS hopes for better integration of the hospital systems and PHKS notes it would be useful to get organ-specific and fetal doses calculated. HUS and OYS both are purchasing a dose management software, which they see as a solution to their current problems in dose monitoring.

Table 7: Results for questions concerning QA in CT. The answers from each hospitals for a specific question are collected

Hospital	**Answer**
What quality assurance test?	
HUS	1) Acceptance tests 2) User phantom test 3) Technical tests included in planned maintenance 4) Annual dose measurements by physicists
Continued on next page	

Table 7 – continued from previous page

Hospital	Answer
TYKS	Check up (including calibration) Clean up (organizes/cleans the system hard drive) Quality (to check the constancy of water and other CT-numbers) Quality constancy Monitors Radiation signal/door lights
PHKS	Daily checkup Weekly CT-number, noise and artifacts. Other mandatory (STUK) tests are done by vendor (Siemens).
OYS	1) Checkup of CT and air calibration of detectors 2) Checkup of HU-values, image noise and artifacts 3) Technical image quality test 4) Movement of patient table 5) Beam width checkup 6) CT dose measurements
How is the test performed?	
HUS	1) Image quality with Catphan phantom and dose production $CTDI$ dose measurement performed by the physicists. Additionally measurements by vendor that are reviewed. 2) The own phantom of the equipment is imaged by axial imaging. Images are sent to web server which analyses the results and forms a summary automatically. 3) Vendor service specifically by their own phantom and meters. 4) Physicists measure annually dose production and compare it to dose display.
TYKS	A phantom is imaged and results analyzed automatically
PHKS	Daily and weekly tests are performed in accordance with the Siemens manual.

Continued on next page

Table 7 – continued from previous page

Hospital	Answer
OYS	1) As CT manufacturer has instructed. Usually shutdown, start, and the machine is then allowed to perform the checkup 2) Vendor specific phantom is imaged, HU- and noise values are compared to reference values and visual inspection for the artifacts is performed 3) Catphan 600 is imaged with hospital specified values and images are analyzed with software (Artiscan). Physicist then compares the results with reference and previous values 4) CT patient table movement is measured/checked according to vendor specification 5) Beam width is checked using Gafchromic films. Film is placed on foam support and axial images are taken with usually used collimation. Physicist then measures the actual beam width from the film and compares it to the used collimation. 6) 16 cm and 32 cm PMMA (acrylic) phantoms are placed individually on textile cradle which is then placed inside the gantry. Dose measurements are performed using 10 cm solid state ionization chamber. Cradle allows physicist to check $CTDI_{vol}$ and DLP values in helical scanning. Axial scan technique is also used.
Who performs the test?	
HUS	1) Vendor mechanics, physicists 2) Radiographer 3) Vendor service 4) Physicists
TYKS	Radiographer
PHKS	Radiographers perform daily and weekly tests.
OYS	1) Radiographer 2) Radiographer 3) Physicist 4) Hospital engineers or vendor engineers 5) Physicist 6) Physicist and sometimes in addition vendor engineers
How often is the test performed?	

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Table 7 – continued from previous page

Hospital	Answer
HUS	1) At the acceptance 2) Once a week with the new scanners. As the performance is confirmed, the test are done once a month. 3) 3-4 times a year 4) Once a year
TYKS	Daily, weekly, or monthly
PHKS	We have daily and weekly tests.
OYS	1) Daily 2) Weekly 3) Biannually 4) Biannually 5) Acceptance test for new equipments and, if necessary, after the x-ray tube has been changed 6) Biannually
Why is the test performed?	
HUS	1) Legislation and standards 2) ST 3.3 by STUK is the foundation for frequency and national quality confirmation for the content 3) Vendor specs and standards 4) Instructions and regulations by STUK
TYKS	STUK or vendor demands
PHKS	The tests are good for monitoring the operations of the medical device. Of course vendor and STUK demand them.
OYS	1) For air calibration and to be sure that the equipment performance is as required. Also vendor and hospital demand/suggest. 2) STUK, vendor and hospital demand 3) STUK and hospital demand 4) STUK, vendor and hospital demand 5) Hospital demand 6) STUK and hospital demand (not sure about the vendor demand)

Table 8: Results for QA

Hospital	**Answer**
Have you noticed difference in quality assurance test methods between different parties?	
HUS	1) With the acceptance tests the vendor part differs between brands. Physicist part is consistent. 2) User tests are based on the own phantoms of the scanners but the analyses based on them are congruent. 3) Technical tests performed during maintenance are vendor specific, though, they are based on IEC standards. The results must fulfill the acceptance criteria by STUK. 4) Done consistently in HUS area.
TYKS	It would be good if the phantoms were the same
PHKS	I do not have experience with other vendors as Siemens. In other modality differences are minor.
OYS	Testing protocols are instructed very specifically so there shouldn't be difference between different test performers.
Do you think the amount of tests used for quality assurance is adequate?	
HUS	Yes. Otherwise we would change the test frequency and quality confirmation program.
TYKS	Yes, but comparison of the results should be easier.
PHKS	I think that the current practice is good. The tests are easy to perform. Nothing serious has gone unnoticed.
OYS	No. At the moment we don't have tools to objectively measure diagnostic image quality, which would be essential for optimal quality assurance program.
What kind of tests could be added in addition to the current ones?	
HUS	Operation test for the tube current modulation. They are relevant for clinical use and optimization.
TYKS	Automatic tests that tell the user the status of the tube
PHKS	-
OYS	See above.
Have you encountered any situations concerning image quality problems where the problems haven't occurred in the tests but appear when imaging a patient?	
HUS	Certain low contrast artefacts are potentially the ones, few of them have been noticed.

Continued on next page

Table 8 – continued from previous page

Hospital	Answer
TYKS	-
PHKS	No.
OYS	Dried up iodine has dropped to the mylar/detector and after the air calibration there has been severe ring artifacts. Some detector elements took wrong conversion values at startup (Daily checkup was performed) and severe hourglass shaped artifacts were seen with the first patient.

Table 9: Results for Patient Dose

Hospital	Answer
How do you monitor patient dose levels?	
HUS	Dose collection software (more systematically in the future) and dose collections with 10 patient samples.
TYKS	DoseWatch or RIS
PHKS	It is easy to calculate the reference levels with GE Dosewatch.
OYS	Normally with $CTDI_{vol}$ and DLP values. In abnormal situations/investigations the effective dose is determined.
What applications and methods do you have in use for patient dose monitoring?	
HUS	Manual collection at the moment, but we are switching to dose collection software around 2018.
TYKS	-
PHKS	We have GE DoseWatch and it contains the necessary tools.
OYS	$CTDI$ and DLP reports can be found from PACS and DLP values are manually written to RIS system. In the near months dose monitoring software is installed at the hospital. For effective dose determination we use CT-Expo and if necessary RPL or MOSFET dose measurements.
Who is in charge of the monitoring?	
HUS	Radiation supervisors and physicists
TYKS	Physicists
PHKS	Medical physicist

Continued on next page

Table 9 – continued from previous page

Hospital	Answer
OYS	Physicist
Which examination types do you have under surveillance?	
HUS	Main indications with reference values, later all examination types in principle.
TYKS	Thx, head, abdomen
PHKS	We can divide material with study code, study description, or used protocol.
OYS	At the moment we only have the ones that have national dose reference levels. After dose monitoring software we aim to have our every protocol under surveillance. We can differentiate different examinations by protocol name and RIS-code.
Do you compare the patient dose levels among different equipments?	
HUS	Yes
TYKS	Yes
PHKS	Yes, we will do so. It's easy, because both devices have the same study codes and protocols.
OYS	Yes, but at the moment comparison is performed only for small amount of protocols (the ones that have DRL's)
How do you compare the patient dose levels between equipments?	
HUS	Distributions, quartiles, means, and scatters
TYKS	With DoseWatch. We aim to harmonize the protocols but some differences are acceptable
PHKS	It is easy with GE DoseWatch
OYS	Mean $CTDI_{vol}$ and DLP values of a small sample size are compared.
How has the acceptable dose level been determined in your hospital?	
HUS	We use our own acceptable levels, that are based on previous results.
TYKS	DoseWatch. STUK DRL's when available and at times local ones
PHKS	We have used the reference levels. Fulfillment of reference levels is not difficult.
OYS	At the moment they do not differ from the given by STUK. After dose monitoring software we aim to.
How do you notice if the dose level has been exceeded or there are abnormalities?	

Continued on next page

Table 9 – continued from previous page

Hospital	**Answer**
HUS	Means and quartiles
TYKS	DoseWatch alerts
PHKS	We use monthly dose report. There I can find if levels are exceeded or there are abnormalities. I get e-mail alert if there is significant overdose.
OYS	At the moment we can't unless the radiographers or radiologist reports those
What kind of solutions would you need to be able to monitor the patient dose levels better?	
HUS	Dose management software
TYKS	Better integration of the hospital systems
PHKS	It might be useful to get organ-specific and fetal doses to be calculated.
OYS	We are about to purchase dose monitoring software which should cover these issues.

Chapter 7

Summary and Conclusions

In this thesis the quality assurance (QA) and patient dose monitoring methods in computed tomography (CT) have been studied. The main objectives are to find out what type of QA testing is performed by the hospitals and how they monitor the patient radiation doses. With QA the goal is to get information on the performed tests by answering the questions: what, how, by whom, how often, and why. Also the differences in methods and practices among different parties, and if the current amount of QA testing is adequate are points of interest. With patient dose monitoring the target is to find out who is responsible for the monitoring, and how the acceptable dose levels have been determined.

In an ideal situation the X-ray beam intensity I falls exponentially with the thickness x of the material it travels. The beam attenuation can be parametrized by linear attenuation coefficient μ when the outgoing beam intensity I can be determined as stated in (2.2). However, actual patient is not homogeneous and the X-ray beam is polychromatic, thus, the intensity I is defined according to (2.5). The X-ray beam energy dependency is not the direct clinical interest so μ is removed by normalizing the tissue CT number relative to that of water. The CT number (in HU) of a tissue at a certain point is defined in (2.7). CT numbers can vary from $-1000\,HU$ for air to $3000\,HU$ for dense bone.

In a CT scan an X-ray beam is generated with an X-ray tube and focused on the patient. During the scan the beam passes trough the patient, while attenuating in the tissue, and attenuation value along each X-ray from the source to the detector is calculated based on the incoming and outgoing intensities. For one slice, CT gathers and digitizes two-dimensional X-ray shadows from several perspectives around the patient. By starting from the obtained data set and working backwards a mathematical reconstruction of the spatial distribution of the X-ray attenuation properties of the materials

responsible for the set of images is acquired. The final result obtained from the image sets and several slices is a three-dimensional image of the patient.

Since CT is based on ionizing X-ray radiation, it is important to monitor patient radiation doses. $CTDI$ is currently the primary metric used to describe the CT radiation output from the scanner. It is useful for specific exam protocols but is not a direct measure of dose. To better take into account the total amount of radiation deposited in the patient the, DLP can be used. However, DLP is not an appropriate risk indicator since it does not take into account the radiosensitivity of the irradiated tissue.

To evaluate the stochastic risk of ionizing radiation exposure effective dose is used. It takes into account the radiosensitivity of the organs and the equivalent doses to all exposed organs. Effective dose is practical for comparing different radiological exposures on common scales but is not an appropriate risk indicator for an individual. To take into account the patient size for dose evaluation $SSDE$ can be used. However, it has some problems and can not be used for organ or effective dose estimation. The most commonly metric referred as patient dose is absorbed dose, which gives the amount of energy deposited per unit mass.

Patient dose is affected by many factors, which include i.e. patient dose, clinical indication, and CT scanning factors. These same factors also affect image quality which is closely related to radiation dose. Image quality is depended on four basic factors which include spatial resolution, noise, contrast, and artifacts. Depending on the diagnostic task, these factors determine the sensitivity and visibility of details.

QA test are divided to acceptance tests and to the ones performed during clinical use. All radiological equipments need to be tested before use with acceptance tests, at specific intervals with periodic testing, after major repairs or maintenance, or if there is a malfunction is suspected. If the operation boundaries of the system are exceeded, proper actions to improve the functions and performance need to be taken. The operation boundaries are divided to acceptability and repair boundaries from which the repair boundaries are stricter.

STUK provides a QA manual which contains the recommended tests for CT scanners (Appendix A). The manual explains the meaning and intervals of the tests and gives guidance on how to perform them. The tests are divided into user and technical tests from which the technical tests are more comprehensive. The core features affecting the system functions are included in both tests. STUK also provides regularly updated acceptability boundaries.

Because of multiple CT vendors and models, numerous CT protocols are available. This makes it hard to keep in track with all used protocols and patient radiation doses. To aid the monitoring, several commercial dose -

tracking software packages are available.

Since QA and patient radiation dose have a central role in the use of CT scanners, it is a point of interest to see how these two areas are handled in the hospitals. This was done by sending pre-selected hospitals a survey which included questions concerning these two topics. The questions concerning QA consisted of two parts: the first one concentrated on the QA tests performed in the hospital and their test setup, and the second one on supporting questions regarding QA tests. The hospitals included in this survey were: The Hospital District of Helsinki and Uusimaa (Helsingin ja Uudenmaan sairaanhoitopiiri, HUS), Turku University Hospital (Turun yliopistollinen keskussairaala, TYKS), Päijät-Häme Central Hospital (Päijät-Hämeen keskussairaala, PHKS), Oulu University Hospital (Oulun yliopistollinen keskussairaala), Kuopio University Hospital (Kuopion yliopistollinen sairaala, KYS), and Seinäjoki Central Hospital (Seinäjoen keskussairaala, SJKS). Even though The Hospital District of Helsinki and Uusimaa consists of several hospitals, the QA and patient dose monitoring methods are centralized, and therefore, it was handled as one unit and referred as one hospital in this thesis.

At the end, answers from four following hospitals were received: HUS, TYKS, PHKS, and OYS. Based on the survey, the hospitals mainly follow the requirements and instructions from STUK and vendors. They perform mostly periodic tests, though, some additional acceptance tests exist. Tests are performed from daily to annual ones. The more frequently (daily and weekly) performed test are performed by the radiographers and other more rare tests by the hospital physicists or engineers. In addition, the vendor engineers perform the more comprehensive technical tests 2 to 4 times a year. The test setups between different parties appear to be consistent, tough, some small variation exists. Since legislation and STUK give very specific guidelines for the tests, there should not be major differences. Hospitals are overall satisfied with the current QA tests. However, some specific tests regarding image quality and X-ray tube are suggested. Certain problems concerning image quality has been encountered in hospitals. These include low-contrast, and ring and hourglass shaped artifacts.

Patient dose monitoring is in half of the hospitals done with a dose management software. The other half is currently using variety of manual methods for the surveillance. All hospitals follow the doses of specific examination types. In the hospitals without a management software manual methods are applied, whereas the other hospitals can do this automatically with the softwares. For the dose acceptance levels, both the ones determined by STUK as well as hospital own values are used depending on the possibilities. The hospitals with dose management software get automatic alerts if the dose

levels are exceeded but the ones without a software have to rely on manual methods and radiologists. All hospitals have a medical physicist in charge of the patient radiation dose monitoring. As for the future improvements hospitals hope for better integration of their systems and certain doses to be calculated. The hospitals currently without dose management software in use are purchasing one in the close future and see this as a solution for their problems.

Based on the results of from the survey the hospitals follow legislation and STUKs regulations well. The amount of QA testing is mostly seen adequate, however, some specific tests could be added and some image quality problems have been encountered despite the tests. To get a better grip on the extent of the these, more hospital should be included in the survey. The questions could also be modified to some extent for acquiring more specific information of the test and problems. In the thesis the idea was also to visit the hospitals and follow how they perform the QA tests. However, due to scheduling issues and the fact that the more specific tests are mainly done biannually or annually, only a couple of visits were paid to the hospitals. To get better picture of the test and their setup, the time period for the further investigations should be long enough to enable the visits. Manual supervision of patient doses is quite time consuming and can be hard to execute in certain situations. Therefore, dose management software is a good option, and is in fact, increasingly the method of choice.

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Appendix A

Quality Assurance Tests

The CT QA tests recommended by STUK can be found here. The user tests are presented in Table 10 and technical tests in Table 11. The tables give the test name or function, reason for the test, and the recommended interval.

Some of the tests recommended for X-ray tubes, generators, image frames, and image displays also apply for CT equipments. The related user tests are presented in Table 12 and technical tests in Table 13.

Table 10: User tests for computed tomography equipments [38].

Name or function	**Reason for the test**	**Interval**
Functionality of the equipment	To check functionality of the equipment based on equipment specification and instructions given in user training.	1 day
CT values	To determine the constancy of water or other reference substance.	1 week
Noise and distortions of an CT image	To determine the noise of an CT image taken of a phantom based on the standard deviation of CT values. To check whether artifacts or other image distortions occur in the image.	1 week
Other tests recommended by the manufacturer		

Table 11: Technical tests for computed tomography equipments [38].

Name or function	**Reason for the test**	**Interval**
Slice positioning range accuracy	To determine correspondence of the position of an axial slice image to adjustment lights and the position of the slice image determined by the scan image.	6 months
CT values	To determine the constancy of water or other reference substance.	6 months
Uniformity of a CT image	To verify the uniformity of a CT image acquired from a phantom by determining the variance of standard deviation of CT values.	6 months
Geometric accuracy of a CT image	To determine the correspondence of the dimensions of a CT image to the actual measures in x and y directions.	6 months
Slice thickness	To determine the slice thickness and compare it to the nominal slice thickness.	6 months (axial imaging), Acceptance test (helical imaging)
Accuracy of patient table position	To determine the longitudinal accuracy (z dimension) of a loaded patient table.	6 months
High contrast resolution	To determine the spatial resolution of a CT imaging based on visibility of high contrast details.	6 months
Contrast threshold or low contrast resolution	To verify the contrast threshold of a CT imaging based on low contrast target resolution.	Acceptance test
Width of dose profile	To verify the cropping of the radiation beam.	Acceptance test
Dose display	To verify the correctness of the dose display at each field size.	6 months 1 year (All collimations)
Other tests recommended by the manufacturer		

Table 12: User tests for X-ray tubes, generators, imaging frames, and image displays that apply for computed tomography equipments. [38]

Name or function	**Reason for the test**	**Interval**
Imaging frame	To verify the functionality of mechanical functions and emergency switches of the X-ray machine, imaging frame, and patient table.	6 months
Functionality of safety devices	To verify the functions of radiation detectors and warning lights, and the state of radiations protectors.	6 months, 1 year for radiation protectors
Operating environment	To check the viewing conditions of image display.	1 week
General visual evaluation	To check the functions of imaging display with test image.	1 week
Other tests recommended by the manufacturer		

Table 13: Technical tests for for X-ray tubes, generators, imaging frames, and image displays that apply for computed tomography equipments. [38]

Name or function	**Reason for the test**	**Interval**
Filtration	To verify the filtration of radiation beam.	Acceptance test, after changing X-ray tube, 1 year with automatic filter changer
Operating environment	To check the viewing conditions of image display.	1 year
General visual evaluation	To check the functions of imaging display with test image.	1 year
Luminosity smoothness	To check the smoothness of luminosity (image brightness) on whole image.	1 year
Other tests recommended by the manufacturer		

Appendix B

Survey for the Hospitals

This appendix includes the survey template sent to the hospitals (see next pages).

Quality Assurance and Patient Dose Monitoring Methods in Computed Tomography

These questions concern my master's thesis about quality assurance and patient dose monitoring methods in computed tomography. The overall goal is to find out how the methods used in both areas differentiate between hospitals. The questions are divided into two: quality assurance and patient dose monitoring.

Hope you can find time to carefully answer to all questions. All additional information that is not asked in the questions is also warmly welcome. Please feel free to contact me if you have any questions concerning the questionnaire or the subject overall.

Thank you in advance!

Quality Assurance *(part 1)*

Here in the first part you can find the questions considering the tests and their setup. The goal of this part is to get a comprehensive idea of the tests you perform in your hospital: What, how, by whom, how often, and why. Please use a table per test so the answers are easier to interpret. You can multiply the table as many times as you need (here is just one for example).

What quality assurance test? *(The name of the test)*	
How is the test performed? *(Please describe briefly the test setup)*	
Who performs the test? (*Hospital physicist / radiographer / other*)	
How often is the test performed? *(Does the cycle differ from instructions?)*	
Why is the test performed? *(Does STUK / vendor / hospital / other demand it to be done?)*	

Quality Assurance *(part 2)*

This second part consists of supporting questions considering quality assurance tests. The goals are to find out if different methods and customs used among different parties, and if the amount of tests is adequate.

Have you noticed difference in quality assurance test methods between different parties? *(Assuming more than one party performs a test with the same goal. What parties? What kind of difference?)*	
Do you think the amount of tests used for quality assurance is adequate? *(Why?)*	
What kind of tests could be added in addition to the current ones? *(Why should they be added?)*	
Have you encountered any situations concerning image quality problems where the problems haven't occurred in the tests but appear when imaging a patient? *(What kind of situations?)*	

Patient Dose Monitoring

In the patient dose monitoring sections the goal is to find out how do you follow patient dose levels, who does it, and how the acceptable levels have been determined.

How do you monitor patient dose levels?	
What applications and methods do you have in use for patient dose monitoring?	
Who is in charge of the monitoring?	
Which examination types do you have under surveillance? *(The dose varies between different examinations. How do you differentiate different examinations?)*	
Do you compare the patient dose levels among different equipments? *(Some equipments may be used only for specific examinations which make comparing difficult.)*	
How do you compare the patient dose levels between equipments?	
How has the acceptable dose level been determined in your hospital? *(Does your level differentiate from STUKs recommendations? Do you have a lot of specific examinations that could affect the acceptable level?)*	
How do you notice if the dose level has been exceeded or there are abnormalities?	
What kind of solutions would you need to be able to monitor the patient dose levels better? *(In addition to the current ones in use. Why these solutions?)*	